THE EFFECT OF TEXTURED SURFACES ON POSTURAL SWAY AND LOWER LIMB MUSCLE ACTIVITY DURING QUIET STANDING IN HEALTHY YOUNG AND OLDER ADULTS

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DECLARATION:

I declare that this thesis is entirely my own work and represents the results of my own research carried out at Teesside University. I declare that no material within this thesis has been used in any other submission for an academic award.

Glossary of Terminology

This glossary clarifies the meaning of terms that will be used frequently throughout this thesis.

Balance – term used to describe the dynamics of body posture, which occur in response to inertial forces acting on the body, in order to achieve a state of equilibrium between the body and the surrounding environment, and prevent the body from falling.

Foot insoles – flat shoe inserts which may be smooth or textured, or comprise mechanical components such as vibratory probes, whose function can include increasing comfort, reducing pain, or enhancing balance performance.

Footwear intervention – term used to collectively define all types of shoe inserts, including foot orthoses, foot insoles, shoes and footwear that includes sandals, unstable shoes and athletic-type shoes.

Foot orthoses – shoe inserts which serve to correct or support optimal function and biomechanical alignment of the foot, and can be custom-made and contoured to individuals' foot shape.

Postural sway – the body's ability to shift when standing upright. Small forward and backward and side-to-side motion of the body is termed postural sway; the magnitude or velocity of which is a measure of balance performance.

Sensorimotor function – simultaneous and coordinated activity between the sensory and motor systems of the body.

Somatosensory – term used to define sensory signals, including light touch, vibration, temperature and joint position sense, which the central nervous system receives from a number of tissues including the skin, muscles, and joints.

ABSTRACT:

This thesis investigated whether different textured surfaces have a role to play in changing balance performance in healthy young and older adults. A review of the literature showed that balance may be improved by standing on surfaces, or wearing foot insoles, with texture compared to those that are smooth, possibly by providing enhanced plantar tactile stimulation. It also showed that textured footwear interventions can influence lower limb muscle activity during functional activities. However, some major gaps in current literature were identified. There was limited evidence relating to the effect of texture in older adults: a population known to show poor balance and at high risk of falling. The importance of the geometric textured pattern in changing balance had not been investigated. It was also unclear whether textured interventions altered lower limb muscle activity, as a component of sensorimotor function of balance control. The purpose of this thesis was to address these important areas of research and contribute novel evidence to the field.

In two separate studies, 24 young and 50 older healthy adults conducted tests of bilateral quiet standing with eyes open and closed on two different textured surfaces and a smooth surface as control. Centre of pressure based sway variables were extracted from a force platform and lower limb muscle activity was collected using surface electromyography, over 30 seconds.

Textured surfaces did not significantly alter sway variables or lower limb muscle activity in healthy young adults. Closer observation of the data tentatively suggested texture may have the capacity to alter anterior-posterior sway and centre of pressure velocity in young adults. These findings supported the aims of this thesis to explore the possibility of a textured effect in people with poor balance ability: older adults.

Textured surfaces significantly changed mediolateral sway range and centre of pressure velocity in healthy older adults, in the absence of visual information. No significant changes were observed for lower limb muscle activity, between the textured conditions. Exploratory sub-group analysis for gender generated speculative data suggesting the effect of texture on balance and muscle activity may be more marked in older females.

Evidence from both studies suggests that relative to control, the two textured conditions have opposite effects on postural sway. There may be an optimal textured pattern which could be therapeutically beneficial for enhancing balance performance in healthy and pathological groups. Further investigation is required.

CHAPTER 1: INTRODUCTION

1.1: Background

Co-ordinated activity between the sensory and motor systems of the body controls upright standing balance (Brooks, 1986). Balance performance can deteriorate when there is a deviation from normal sensorimotor function, be it through acute injury, chronic pathology or the normal ageing process (Bonfim et al., 2008, Laughton et al., 2003, McKeon and Hertel, 2008). Age-related physiological changes can lead to degradation of postural control mechanisms, with functional implications being postural instability and a risk of falling (Lord et al., 1994).

Footwear interventions, including modified athletic and standard shoes, foot insoles and foot orthoses have been explored as potential devices to improve functional performance in healthy and pathological groups. Footwear interventions may bring about their effects by influencing biomechanical alignment, motion control or facilitating shock absorbance. However, more recently, enhanced sensory input to the plantar surface of the feet has been considered as a factor in footwear interventions (Hertel et al., 2005, Mundermann et al., 2006, Nurse et al., 2005).

Evidence is emerging concerning the use of textured floor surfaces or foot insoles, as a means to improve balance performance in young, middle-aged and older adults. The term ‘texture’ refers to the upper surface of flooring or footwear interventions comprising indentations, which are hypothesised to increase sensory afferent feedback to plantar cutaneous mechanoreceptors. To the author’s knowledge, only four studies have previously investigated the effect of a textured surface or foot insole on balance performance in adults (Corbin et al., 2007, Palluel et al., 2008, Watanabe and Okubo, 1981, Wilson et al., 2008), with only one research group exploring a population of healthy older people (Palluel et al., 2008). Thus, there is a large gap in current literature concerning the effect of texture on balance performance in older adults: those at risk of impaired postural control due to age-related physiological changes.

Two further studies have concluded that textured insoles can significantly alter lower limb muscle activity, during walking (Kelleher et al., 2010, Nurse et al., 2005).

The current thesis is concerned with modifying the interface between the plantar surface of the foot and the ground, by way of introducing a textured surface. This concept of using texture as a medium to enhance plantar tactile stimulation for balance control will be explored in healthy young and older adults in the first instance.

The current thesis is unique, as to date, no study has explored the effect of textured surfaces on the dual-measurement of postural sway and lower limb muscle activity simultaneously in healthy young or older adults, and thus the effects of texture on sensorimotor function. This research is also the first to investigate two different textured surfaces and identify whether an optimal geometric textured pattern exists for improving balance performance. This thesis will generate vital evidence relating to the future potential for textured surfaces to be used as domestic flooring or footwear interventions, for improving balance performance in adults.

1.2: Thesis Structure

This thesis is structured in a way that leads the reader through two separate studies:

Chapter 2 discusses current literature surrounding normal and impaired sensorimotor control of standing balance and the use of footwear interventions to enhance balance, including those that are textured.

Chapter 3 presents a literature review of the methodology used in this thesis, including the instrumentation and test procedures.

Chapters 4-6 report the background, methods, results, and discusses the findings from healthy young adults. This study explores the effect of textured surfaces on postural sway and lower limb muscle activity during quiet standing, in healthy young adults. Within-session reliability for measures of postural sway and lower limb muscle activity in healthy young adults are also presented.

Chapters 7-9 report the background, methods, results, and discusses the findings from healthy older adults. This is a study of the effect of textured surfaces on postural sway and lower limb muscle activity during quiet standing in healthy older adults.

Chapter 10 discusses the overall findings for the effect of texture on balance performance and muscle activity in healthy young and older adults. Study strengths, limitations and overall conclusions are presented, along with recommendations for future research.

CHAPTER 2: BACKGROUND AND LITERATURE REVIEW

2.1: Introduction

This chapter will discuss sensorimotor control of standing balance in health, injury, pathology and ageing. General interventions used to enhance balance performance, including footwear interventions will be discussed. A detailed review of current evidence surrounding the effect of textured surfaces and foot insoles on balance and muscle activation, and their proposed underlying mechanisms, will be presented. The chapter will conclude with the aims of this thesis.

2.2: Sensorimotor Function and Standing Balance

Exploring the role of the sensorimotor system in balance control in healthy young adults is important for understanding how pathological and age-related physiological changes can lead to postural instability. Upright standing balance is controlled by the central nervous system (CNS), which integrates visual, vestibular and somatosensory information about body position, to initiate appropriate muscular responses at the trunk and lower limbs (Brooks, 1986). Therefore, regulation of motor responses is dependent on incoming information from afferent neurons, which possess sensory receptors at their peripheral ending, such as plantar cutaneous mechanoreceptors (Shaffer and Harrison, 2007).

The musculoskeletal system controls joint and body segment alignment when standing (Fitzpatrick et al., 1994), yet postural adjustments occur in response to sensory information. Higher level neurological processes integrate sensory and musculoskeletal components, leading to anticipatory or adaptive postural responses (Shumway-Cook and Woollacott, 1995).

2.2.1: Sensory contributions to balance

As an individual’s surrounding environment constantly changes, so too does the weighting of dependence on each sub-sensory system. When standing on a firm, fixed surface, in well-lit surroundings, healthy adults predominantly rely on somatosensory information for balance control (Horak, 2006, Peterka, 2002, Shumway-Cook and Horak, 1986). When the supporting surface is unstable or compliant, visual inputs predominate (Peterka and Loughlin, 2004, Peterka, 2002). Vestibular information regulates balance when visual and somatosensory inputs are conflicting (Nashner, 1982). This section will highlight current evidence relating to the role of somatosensory information for balance control.

Experimentally-induced suppression (Eils et al., 2004, Eils et al., 2002, Magnusson et al., 1990) and augmentation (Vaillant et al., 2008) of somatosensory information has been shown to significantly affect postural stability. Immersing the feet in ice is a common method used to experimentally induce hypoanaesthesia, or in simple terms reduce plantar cutaneous sensitivity, and thereafter investigate balance performance relative to normal thermal conditions. Using such procedures to dampen plantar sensory information in healthy adults has been shown to significantly increase anterior-posterior (AP) sway during quiet standing with eyes open (Magnusson et al., 1990) and lead to cautious walking patterns, defined by significant reductions in peak braking and acceleration forces at initial foot contact (Eils et al., 2004), and peak pressure under the heels and toes (Eils et al., 2002). Increased AP sway when standing quietly is clinically undesirable, suggesting the centre of mass (CoM) is moving closer to the limits of stability, which if exceeded, can result in loss of balance. Current understanding of the clinical relevance of changes in sway parameters is discussed further in chapter 3, sections 3.2 and 3.3. The clinical relevance of reduced braking and acceleration forces during walking may translate to a reduced capacity for the body to execute effective corrective postural responses when dynamic balance is threatened, increasing the risk of falling. Alterations in gait patterns were also accompanied by significant reductions in the amplitude of lower limb muscle activity, during various stages of the gait cycle (Eils et al., 2004). This evidence clearly supports the role of somatosensory information for static and dynamic balance control. However, attributing observed changes in postural sway and gait solely to suppressed sensory receptor activity must be interpreted with caution, as ice immersion may also alter intrinsic foot muscle function and nerve conduction.

Performing manual massage and mobilisations at the feet and ankles to enhance somatosensory feedback, has shown to assist balance control in older adults, when standing with eyes closed (Vaillant et al., 2008). Immediately upon closing the eyes, postural sway parameters were observed to increase in both control and intervention conditions. However, following a 10 second adaption period to ‘no vision’, therapeutic manipulation prevented further increases in AP and mediolateral (ML) centre of pressure (CoP) movement (Vaillant et al., 2008). Under control conditions, sway measures continued to increase after the adaptation period. The clinical relevance of this change in standing balance is that a reduction in the magnitude of postural sway indicates that the CoM is moving over a shorter distance and is less likely to reach or exceed the limits of stability, thus the body is better balanced. This evidence suggests enhanced tactile plantar stimulation may have the capacity to improve balance and compensate for suppressed visual inputs (Vaillant et al., 2008). Similar evidence exists relating to the effect of texture on postural sway measures during upright standing with eyes closed (Corbin et al., 2007, Palluel et al., 2008, Watanabe and Okubo, 1981).

2.2.2: Muscular contributions to balance

When standing quietly, the CNS uses proprioceptive and sensory information to regulate muscle tone (Hayashi et al., 1988), so that it is proportional to gravitational and inertial forces acting upon the body. An appropriate level of lower limb muscle activity must be set in response, or prior, to fluctuations in postural sway (Laughton et al., 2003).

Winter et al. (1998) proposed that all postural adjustments are controlled by the ankle muscles, which act like springs, restoring the CoM to a central position (Winter et al., 1998). However, during quiet standing, joint movement occurs throughout the body (Day et al., 1993), together with sporadic lower limb muscle activity (Loram et al., 2005b). Intermittent activity between the plantarflexors and dorsiflexors will cause the CoP to move in the AP direction. Similarly, ML CoP movement will occur through changes in gluteal, hip abductor and adductor and peroneal muscle activity (Day et al., 1993).

The postulation by Winter et al. (1998), suggesting the body should be modelled as an inverted pendulum, with the ankle being the only moveable joint, has been questioned by other research groups. Loram et al. (2005a) disagree that the calf muscles function like springs. Studies have been conducted, tracking changes in the length and activity of soleus and medial gastrocnemius, during quiet standing (Loram et al., 2005a). Anterior body sway caused both muscles to shorten, asynchronously, and not lengthen as suggested previously (Winter et al., 1998). Soleus and medial gastrocnemius then lengthen as the body sways posteriorly, back to vertical (Loram et al., 2005a). Thus, it is possible upright posture may be coordinated through peripheral feedback or higher level control mechanisms, which guides the CNS to appropriately adjust muscle length when necessary (Loram et al., 2005a). This evidence also questions the role of stretch reflexes for balance control, which occur in response to muscle elongation (Levy, 1963).

Evidence by Masani et al. (2003) supports the notion that lower limb muscle activity for balance control is governed by higher CNS processes. This research group suggested calf muscle activity may be anticipated prior to changes in body position. Gatev et al. (1999) also reported lateral gastrocnemius activity preceded CoM movement. Masani et al. (2003) proposed that plantarflexor muscle activity is determined from information relating to CoM velocity. As CoM velocity cannot be measured directly by any sensory system, the CNS may integrate multi-sensory inputs to detect this movement. Plantar cutaneous mechanoreceptors may provide vital information relating to changes in the position and acceleration of foot pressure and therefore CoM movement. Thus, the findings of Masani et al. (2003) and Gatev et al. (1999) both suggest lower limb muscle activity occurs through feed-forward mechanisms and in anticipation of CoM movement.

2.3: Balance Performance in Health

Demographic, physiological and lifestyle factors can influence upright balance control in healthy adults. Rogind et al. (2003) set out to define reference data for postural sway measures in healthy adults aged 20-70 years, considering gender, age, body mass, joint movement, smoking history and alcohol consumption. Increased postural sway was only significantly correlated to age and the difficulty of the balance task (Rogind et al., 2003).

Maintaining a healthy lifestyle, which includes exercise participation, has shown to enhance functional balance performance in young (Gerbino et al., 2007), middle-aged (Thornton et al., 2004) and older adults (Brooke-Wavell and Cooling, 2008). Healthy, young, female dancers performed significantly better in 5 out of 10 standing balance tests, compared to female soccer players (Gerbino et al., 2007). This may suggest participation in sports demanding high levels of balance control have transferrable training effects to functional activities. Previously sedentary middle-aged females have shown significant improvements in dynamic balance performance, compared to age-matched controls, upon completing a programme of Tai Chi classes (Thornton et al., 2004). Long-term Tai Chi participation has been shown to reduce postural sway during single-limb stance and increase balance confidence in healthy older adults (Tsang and Hui-Chan, 2005). Older females, who regularly participated in lawn bowls, were reported to have better postural stability when standing on a compliant surface, than sedentary age-matched controls (Brooke-Wavell and Cooling, 2008). Thus it appears exercise can enhance or preserve balance control in healthy adults.

2.4: Balance Performance and Injury

Acute musculoskeletal injury can impair postural control (Bonfim et al., 2008). Young adults with unilateral anterior cruciate ligament (ACL) injury have shown greater amplitude of AP and ML sway when standing on one leg, compared to healthy controls (Bonfim et al., 2008). Such declines in balance performance may be due to loss of sensory information from the ACL, which is important for postural control. From a clinical perspective, the severity of ACL sensory loss may determine the magnitude to which CoP movement undesirably increases. Should upright balance be lost, impaired proprioception at the knee may also reduce the capacity to execute effective corrective postural responses, thus resulting in a fall. When sensory information was enhanced by allowing light fingertip touching on a stationary bar, both control and injured groups showed improvements in postural sway measures. However, this effect was more marked in those with ACL injury (Bonfim et al., 2008).

Chronic musculoskeletal conditions including low back pain (Mann et al., 2009, Mientjes and Frank, 1999), osteoarthritis (Hassan et al., 2001, Hinman et al., 2002), rheumatoid arthritis (Tjon A Hen et al., 2000) and chronic ankle instability (McKeon and Hertel, 2008) can impair postural control. Adults with chronic low back pain have shown deficits in upright standing balance, compared to healthy controls (Mientjes and Frank, 1999). Significant differences in ML sway were observed between pathological and healthy groups when standing balance was sufficiently challenged, through dual-tasking or the removal of visual information (Mientjes and Frank, 1999). Young females with chronic low back pain have also shown greater AP and ML sway during quiet standing compared to healthy controls (Mann et al., 2009). Significant differences in CoP velocity were identified between the two groups, only when standing with eyes closed. Therefore, if low back pain can increase both the amplitude and velocity of CoP movement, then this group of patients may be at high risk of falling. The clinical relevance of this evidence suggests that patients with low back pain may show deficits in maintaining their CoM within the safe limits of their base of support and also have less time to recover balance should they experience a de-stabilising perturbation. This evidence also suggests that balance impairments, originating from chronic injury, appear to be more marked when postural stability is substantially threatened (Mann et al., 2009).

Adults presenting symptomatic knee osteoarthritis have shown significantly greater ML postural sway during quiet standing on a firm surface with eyes open, than healthy controls (Hinman et al., 2002). When standing on a firm surface with eyes closed, AP and total sway measures were significantly greater in the pathological group (Hinman et al., 2002). Similarly, during a dynamic step test, osteoarthritic participants took fewer steps within an allotted time period, indicating poorer balance ability than the control group. From a clinical perspective, this evidence may suggest that osteoarthritis could have undesirable effects on the ability to execute corrective stepping reactions to prevent falling. Impaired quadriceps strength and activation, coupled with a loss of knee joint proprioception are changes reported with knee osteoarthritis (Hurley et al., 1998), which may contribute to declines in balance ability.

Hassan et al. (2001) provide similar evidence, indicating that older adults with symptomatic knee osteoarthritis show greater lateral postural sway, compared to healthy controls during quiet bipedal standing with eyes closed. In older adults with rheumatoid arthritis and severe knee joint impairment, measures of AP and ML CoP velocity during quiet standing were reported to be on average 80% greater than healthy controls (Tjon A Hen et al., 2000). Thus, with the CoP moving at a faster velocity, older adults with lower limb arthritic conditions may have insufficient time to respond to postural disturbances, increasing their likelihood of falling.

Young adults presenting chronic ankle instability have shown deficits in postural control, during single limb stance with eyes closed, in comparison to healthy adults (McKeon and Hertel, 2008). Thus ankle injury appears to impair balance control (McKeon and Hertel, 2008, Wikstrom et al., 2010). Musculo-tendinous changes around the ankle or loss of proprioceptive information may contribute to changes in postural control mechanisms after injury.

2.5: Balance Performance and Ageing

There is general consensus throughout current literature that measures of postural sway increase with age (Benjuya et al., 2004, Berger et al., 2005, Era et al., 2006, Laughton et al., 2003, Raymakers et al., 2005), representing a gradual deterioration in balance performance. Differences in sway between age groups became more prominent the greater the balance challenge (Era et al., 2006).

2.5.1: Gender differences in balance performance in older adults

Current evidence suggests a possible gender difference in balance ability in older adults. Older women have shown significantly greater AP and total maximal centre of gravity movement compared to males, during bilateral standing whilst simultaneously throwing a ball overhead (Kinney La Pier et al., 1997). No significant gender differences have been shown during less challenging balance tests, including quiet standing. Thus, older females appear to have poorer postural control during dynamic balance challenges. Gbiri and Fabunmi (2006) report similar findings to Kinney La Pier et al. (1997). Healthy, older males scored significantly higher than females during static (Sharpened Romberg Test) and dynamic (Functional Reach Test) balance tests: where higher scores represented better balance (Gbiri and Fabunmi, 2006). However, other studies report that older males show greater postural sway than females (Era et al., 1997, Masui et al., 2005), or there are no significant gender differences in balance measures in older people (Bryant et al., 2005).

2.5.2: Characteristic changes in postural sway patterns with age

There are a number of key changes in postural sway characteristics which can indicate when the body is unstable during quiet standing. With increasing age, postural sway patterns can become less complex, showing loss of variability in CoP movement (Van Wegen et al., 2002). This could result from reduced exploratory postural activity (Van Emmerik and Van Wegen, 2002) and contribute to functional decline (Lipsitz, 2002). Reduced sway variability may exacerbate postural instability, as there are fewer solutions available to make postural corrections. When the body becomes less rigid, it is more adaptable and capable of executing compensatory balance reactions when stability is challenged (Van Emmerik and Van Wegen, 2002). Applying sub-sensory mechanical noise to the plantar surface of the feet in older people prone to falling has been shown to improve postural stability, generating more complex sway patterns (Costa et al., 2007). Therefore, limited exploratory behaviour of the CoP may represent an inability to cope with balance demands.

Other research groups suggest ageing can lead to increased variance, amplitude and velocity of sway (Baloh et al., 1998, Benjuya et al., 2004, Berger et al., 2005, Laughton et al., 2003) particularly in the ML direction (Berger et al., 2005, McClenaghan et al., 1995, Raymakers et al., 2005); with such changes being detrimental to balance performance. Greater lateral CoP movement is associated with a heightened risk of falling in older adults (Maki et al., 1994). Significant differences have been reported for ML sway variance and root mean square between old (62-75yrs) and very old (75-96yrs) people (Berger et al., 2005). Stabilisation of lateral sway may also be delayed in older people, due to diminished sensory information, causing larger magnitudes of sway (Berger et al., 2005). Age-related changes in postural sway characteristics are not exclusive to lateral balance measures. Other research suggests ageing can lead to increases of a similar magnitude in both AP and ML sway amplitude (Baloh et al., 1998).

2.5.3: Age-related physiological changes and balance performance

Deterioration of postural control with increasing age, may be attributed to sensorimotor changes, including impaired foot sensation, visual acuity, vestibular function, muscle strength and reaction times (Lord et al., 1991, Lord et al., 1994). Within the context of the current thesis, it is important to understand how age-related changes in the somatosensory system and muscle function may impair balance control.

Ageing can lead to reductions in the number and size of axons, and deterioration of surrounding myelin sheaths (Menz, 2008). Loss of nerve fibres will reduce the amplitude of nerve impulses transmitted, whilst loss of myelin will affect nerve conduction velocity (Dorfman and Bosley, 1979, Hurley et al., 1998). These changes occur predominantly in the longest axons of the body: those extending from the spinal cord to the lower limbs (Olney, 1985). This is of relevance to the current thesis as peripheral sensory system decline may affect communication between plantar cutaneous mechanoreceptors and the CNS.

Older adults show reduced density and sensitivity of cutaneous mechanoreceptors, rigid and inelastic dermal tissue and peripheral nerve degeneration (Kenshalo, 1986). Such structural changes in peripheral sensory receptors can be identified through poor performance in clinical assessments of tactile sensitivity, spatial acuity and vibration sense at the feet and ankles (Menz, 2008). As proprioceptive and vibration sense provide vital information about a supporting surface, these changes may have serious implications on postural stability in older people. Plantar tactile sensitivity is one of several independent predictors of balance and functional performance in older people (Menz et al., 2005). Diminished proprioception and sensation in the lower limbs is associated with multiple falls (Lord et al., 1994), and an increased likelihood of falling (Judge et al., 1995) in older people. It remains unknown whether applying enhanced plantar tactile stimulation would be of benefit to older people and surpass changes in receptor sensitivity.

If age-related postural instability originates from dampened or slow transmission and processing of sensory information (Richardson et al., 1996), then interventions hypothesised to enhance sensory feedback, such as textured surfaces, will have limited influence on motor responses. However, textured surfaces could serve to amplify weak sensory input at the feet and thus provide vital information about the supporting surface.

Older adults are reported to show the following age-related muscular changes: a reduction in the number of functioning motor units (Porter et al., 1995), increased time to peak tension, particularly in plantarflexor and dorsiflexor muscles (Vandervoort and McComas, 1986), and reductions in the size of fast twitch muscle fibres (Lexell et al., 1988) which may contribute to the slowing of muscular contractions and force development (Lexell et al., 1988). Stelmach et al. (1989) and Dixon et al. (2004) also report slow reflex responses. These changes may influence the ability to initiate postural responses.

Postural muscle activity, including onset timing and sequencing, significantly differs between young and older adults (Woollacott and Manchester, 1993). Fewer older adults showed preparatory postural muscle activity in erector spinae and hamstrings before raising their arm during upright standing, compared to young adults (Woollacott and Manchester, 1993). However, more frequent quadriceps activity, observed in the older adults, was suggested to represent excessive caution or misjudgement regarding the extent to which arm movement would disturb postural control (Woollacott and Manchester, 1993). In healthy older adults, lower limb muscles have been shown to remain active for a longer duration during quiet standing, compared to young adults (Laughton et al., 2003). This is particularly true for vastus lateralis and antagonistic activity between the quadriceps and hamstrings. However, no significant differences were reported between young and older adults for simultaneous measures of postural sway. Therefore, this evidence (Laughton et al., 2003, Woollacott and Manchester, 1993) suggests young and older adults use different postural strategies to maintain upright stance.

Greater lower limb muscle activity in older adults can result from multiple factors. The stooped, rigid posture commonly observed with old age undesirably shifts the CoM anteriorly. Therefore, increased hamstring activity will move the CoM more posteriorly and enhance stability (Accornero et al., 1997). Prolonged quadriceps activity may compensate for underlying weakness, commonly reported in vastus lateralis (Hurley et al., 1998). In contrast, greater lower limb muscle activity may serve to augment distal joint proprioception (Cordo et al., 1996).

Research suggests there is an association between sway variability and age-related contractile changes in calf muscles (Onambele et al., 2006). Increases in CoP movement during single-limb and tandem standing have shown to be significantly associated with decreased plantarflexor strength, size and activation capacity. Such observations were not made during bilateral standing, suggesting the influence of age-related musculoskeletal changes may only be apparent in non-functional or unfamiliar postures (Onambele et al., 2006).

Young healthy adults appear to use pre-synaptic mechanisms to regulate motor control. As this system is less effective in older adults (Earles et al., 2001) increased lower limb muscle activity and co-activation may provide an alternative mechanism (Earles et al., 2001, Laughton et al., 2003, Woollacott and Manchester, 1993).

2.6: General Interventions to Enhance Balance Performance

Interventions to enhance balance ability in young and older adults are commonly used to optimise sports performance through prophylactic or rehabilitative mechanisms, or reduce falls risk in older populations. Desirable outcomes include reductions in the magnitude and velocity of CoP movement which can translate to better functional ability. Such interventions include exercise prescription including core stability (Kaji et al., 2010), aquatic exercise (Suomi and Koceja, 2000), or land-based exercise programmes to enhance muscle strength, flexibility, coordination, balance and endurance (Campbell et al., 1997, Judge et al., 1993, Lord et al., 1995). Specific balance-enhancing exercise classes can significantly improve functional performance in older people (Nagy et al., 2007). Some interventions have been shown to improve balance performance by modifying human biomechanics or the interaction between the body and external environment. These include neuromuscular taping (Hassan et al., 2002), modified footwear (Menant et al., 2008a, Menant et al., 2008b), custom-made foot orthoses (Mattacola et al., 2007, Olmsted and Hertel, 2004) or assistive devices such as walking canes (Bateni and Maki, 2005).

2.7: Footwear Interventions

Studies investigating the influence of midsole hardness in athletic footwear have shown that a softer midsole can lead to postural destabilisation (Perry et al., 2007, Robbins et al., 1994). Similarly in older adults, a harder shoe midsole may have the potential to enhance balance performance (Menant et al., 2008a). A softer interface between the foot and supporting surface may provide less mechanical support and insulate sensory information from plantar mechanoreceptors (Perry et al., 2007). This could increase both mechanical and sensory demands for maintaining upright standing balance.

Heel height is also an important shoe characteristic in balance control. A review by Cowley et al. (2009) highlights the detrimental effect of wearing shoes with high heels on standing balance. Even minimal heel elevation can lead to postural instability in older adults (Menant et al., 2008a).

Custom-made foot orthoses can enhance balance performance in individuals with high- or low-arched feet. Semi-rigid foot orthoses have shown to improve postural stability during quiet standing in healthy young adults with high-arched feet (Olmsted and Hertel, 2004), whilst rigid foot orthoses can specifically reduce ML sway in those with low-arched feet (Rome and Brown, 2004). Evidence also suggests foot orthoses may be useful for enhancing balance performance when the lower leg muscles are fatigued (Oschendorf et al., 2000).

The effects of vibrating (Galica et al., 2009, Hijmans et al., 2007b, Hijmans et al., 2008, Novak and Novak, 2006, Priplata et al., 2003), mineral (Masse et al., 2000), magnetic (Suomi and Koceja, 2001) and dynamic (Ramdharry et al., 2006) foot insoles on postural stability have been explored. The relevance of these studies to the current thesis centres around postulations that improved balance performance when wearing such footwear interventions may be attributed to alterations in sensory information at the plantar surface of the feet. However, as these are not textured footwear interventions, the studies will only be outlined briefly.

Vibrating insoles, which apply sub-sensory mechanical noise to the plantar surface of the feet, have been shown to significantly; reduce postural sway in healthy young and older adults (Priplata et al., 2003), improve standing balance in adults with neuropathy (Hijmans et al., 2008), alter gait parameters in individuals with Parkinson's disease (Novak and Novak, 2006), and reduce gait variability in older fallers (Galica et al., 2009). Vibratory feedback may lead to earlier detection of changes in foot pressure, thus earlier corrective postural responses (Hijmans et al., 2007a).

Flat insoles containing powdered mineral derivatives, have been shown to significantly improve asymmetries in postural alignment during quiet standing, compared to control conditions (Masse et al., 2000). Whilst these authors did not directly measure balance performance, changes in body posture, detected at the foot, may induce alterations in postural sway.

Wearing magnetic insoles has shown to significantly reduce postural sway in adults. However, such changes may occur due to a combination of mechanisms including altered blood flow and sensory information at the feet (Suomi and Koceja, 2001).

Dynamic foot orthoses, contoured to the foot arches of individuals with neuromuscular conditions such as Multiple Sclerosis, can significantly increase CoP velocity and move the mean position of the CoP, during quiet standing (Ramdharry et al., 2006). This effect is more marked with eyes closed. The contoured metatarsal pad located in the mid-foot, may have contributed to declines in postural stability, by emphasising stimulation of mechanoreceptors in the mid-foot region (Ramdharry et al., 2006). These mechanoreceptors are considered less important for detecting dynamic changes in posture, compared to fore- and rear foot receptors (Chiang and Wu, 1997, Meyer et al., 2004). However, long-term wear of the dynamic foot orthoses significantly reduced postural sway during quiet standing with eyes closed, with or without the orthoses in situ. Therefore, contoured orthoses may initially induce de-stabilisation, followed by a functionally beneficial training effect.

A balance-enhancing foot insole was recently developed by Maki et al. (2008), following studies exploring the effect of enhancing sensation around the plantar surface borders of the feet (Maki et al., 1999). A compliant raised ridge of polyethylene tubing (3mm outer diameter and 1mm inner diameter) was fixed around the posterior perimeter of the rearfoot. The authors claimed this ridge would indent the skin, stimulating mechanoreceptors around the periphery of the sole of the foot when the CoM moves away from the equilibrium point and individuals start losing their balance (Maki et al., 2008). The raised ridge was constructed from elastomeric material to ensure stimulation of mechanoreceptors occurred only when the CoM moved within close proximity to the limits of the base of support, thus preventing habituation to the sensory stimulus (Maki et al., 2008)

Studies have been conducted to explore the long-term effects of the balance-enhancing insole on dynamic postural stability in older adults with mild age-related loss of foot sensation (Maki et al., 2008, Perry et al., 2008). Older adults walked over uneven terrain (simulated using inclined platforms) with and without balance-enhancing insoles. Lateral balance control was calculated by measuring the lateral stability margin. This parameter considers the lateral distance between the CoM and boundary of the base of support (Perry et al., 2008), with larger values indicating greater lateral stability. After wearing the insoles for 12 weeks, lateral stability was significantly greater with the balance-enhancing insoles when stepping on laterally and anteriorly inclined platforms (Perry et al., 2008), compared to control. This significant effect was observed both pre- and post-intervention, indicating habituation to the plantar sensory stimulus had not occurred. This is a clinically important finding which suggests that the balance-enhancing insoles may improve older people's ability to maintain balance in conditions simulating day to day life, such as uneven pavements.

The balance-enhancing insole has also shown to significantly increase single-limb support time during level ground walking in individuals with Parkinson's disease, relative to a flat insole (Jenkins et al., 2009). The maximum effect size was an increase of 6.4ms between insole conditions, occurring at gait trial 4 of 5. It is questionable whether this short duration is clinically significant. However, it appears that the balance-enhancing insole has the capacity to 'buy' more time for stabilisation during walking, where the location of the CoM is constantly changing.

Considering the construction of the balance-enhancing insole, the flexible raised ridge extends from the first to the fifth metatarsal heads. Therefore, as this tubing essentially 'cups' and provides a barrier around the heel and lateral periphery of the foot, it is not unexpected that this insole has shown to improve lateral stability (Maki et al., 2008, Perry et al., 2008) and single limb support time (Jenkins et al., 2009). This may suggest the balance-enhancing insole brings about its effect on postural stability through mechanical, rather than sensory, mechanisms. The balance-enhancing insole may only stimulate cutaneous mechanoreceptors located near the periphery of the foot, thus its design does not conform with the concept of total and even plantar stimulation, hypothesised by Watanabe and Okubo (1981). As will be discussed later in this chapter, when the whole plantar surface of the foot receives stimulation from a textured surface, improvements have been observed in both AP and ML sway measures (Watanabe and Okubo, 1981).

Common to all these footwear interventions, is the inability to determine whether their underlying mechanism is mechanical or sensory in nature, and whether they work on a similar principle to textured surfaces or foot insoles.

Recent evidence suggests foot orthoses may also have a significant effect on lower limb muscle activation patterns. A critical review published by Hatton et al. (2008) (Appendix 1), examines the evidence for changes in muscle activation patterns when wearing foot orthoses, and explores the proposed mechanisms by which foot orthoses may bring about changes in lower limb muscle activity. It is proposed that different mechanisms may occur by which foot orthoses affect muscle activity, due to their differing construction and design.

Custom-made foot orthoses have shown to significantly increase the intensity of lower limb muscle activity, particularly in high-frequency bands, during the stance phase of running (Mundermann et al., 2006). This evidence supports the concept that changes in muscle activation, due to altered plantar sensory input, may be influenced by the intensity of the task. As running leads to an electromyogram with high-frequency content, due to the activation of fast twitch fibres (Wakeling et al., 2002), the foot orthoses investigated by Mundermann et al. (2006) may have primarily influenced faster motor units. However, running also leads to muscle vibration as the foot impacts the ground, requiring immediate muscle activation to dampen this vibration. Therefore, mechanisms underlying the effect of these foot orthoses on muscle activity could be mechanical or sensory in nature.

Wearing therapeutic shoes during walking has been shown to significantly alter tibialis anterior activity, relative to control conditions. Upon initial foot contact, tibialis anterior activity significantly decreased, followed by a significant increase in activity during the swing phase of gait (Romkes et al., 2006). This evidence from Romkes et al. (2006) may compliment the findings of Tomaro and Burdett (1993) who observed a significant increase in the duration of tibialis anterior activity when walking with semi-rigid foot orthoses. Footwear interventions, including therapeutic shoes and foot orthoses, may be capable of reducing lower limb muscle activity (Romkes et al., 2006), equating to greater fatigue resistance (Tomaro and Burdett, 1993).

2.8: Texture and Postural Stability

A small number of studies have investigated the use of texture, as a floor surface or foot insole, to improve postural stability during quiet standing in adults (Corbin et al., 2007, Palluel et al., 2008, Watanabe and Okubo, 1981, Wilson et al., 2008). It is worth noting that most of these studies were published after the research in this thesis had commenced. Textured surfaces and footwear interventions which have been investigated include; laboratory-prepared stimulating surfaces (Watanabe and Okubo, 1981), off-the-shelf footwear products (Palluel et al., 2008), and foot insoles cut from commercially available textured sheeting (Corbin et al., 2007, Wilson et al., 2008). With no consensus in the geometry of the textured pattern, density or flexibility of the material, depth or spacing between indenting protrusions, such factors could result in differing degrees of plantar stimulation between the studies, limiting the ability to compare findings.

Watanabe and Okubo (1981) investigated the effect of stimulating plantar mechanoreceptors, using a textured surface, on postural stability in 20 healthy, young male adults (age mean 24.5 years). Their textured surface comprised regularly spaced shotgun balls of 20mm diameter, protruding approximately 1mm from the surface (Table 2.1). Three textured experimental conditions were defined by the distance between the balls including; 10mm, 15mm and 20mm. During quiet standing with eyes closed over 20 seconds, shotgun balls placed 10mm apart, significantly reduced sway area by 25mm^2, and total sway length by 46.8mm, compared to control. Shotgun balls placed 15mm apart showed similar reductions relative to the control condition: 29mm^2 (sway area) and 45mm (total sway length) (Watanabe and Okubo, 1981). Whilst the shotgun balls placed 20mm apart increased sway area by 10mm^2, total sway length was observed to decrease by 21mm, compared to baseline. The magnitude of this effect was not statistically significant and less than that when the balls were placed closer together.

This evidence suggests there may be an optimal centre-to-centre distance between protrusions, which will achieve greatest plantar stimulation. With more protrusions contacting the feet (per unit square area) in the 10mm- and 15mm-spaced shotgun conditions, Watanabe and Okubo (1981) reported increased tibial nerve discharge rate. What remains unknown is how changes in the rate of discharge from plantar mechanoreceptors contribute to alterations in body sway.

The findings from Watanabe and Okubo (1981) suggest optimal plantar stimulation can significantly assist balance control in the absence of visual sensory information. However, all three shotgun ball conditions also reduced sway area, relative to control, during quiet standing with eyes open: 29mm^2 (10mm spacing), 23mm^2 (15mm spacing) and 15mm^2 (20mm spacing). This may imply using plantar stimulation to improve balance is not dependent on the redundancy of other sensory information, but rather the construction, including the shape, centre-to-centre distance and depth of protrusions, comprising the stimulating interface between the foot and ground. However, the clinical relevance of the findings from Watanabe and Okubo (1981) is unclear. Total sway length is defined as the total distance travelled by the CoP over the duration of a trial. Current understanding is that an increase in total sway length indicates a reduction in stability. This may not be clinically useful information as large sway length could also occur in stable conditions.

As one of the earliest studies investigating the influence of texture on postural stability, sway data was extracted from a system comprising a platform, goniometer and pen recorder (Watanabe and Okubo, 1981). This apparatus lacks the comprehensive precision of current day piezoelectric force platforms, thus challenging the reliability of their findings (Watanabe and Okubo, 1981), and comparability with later investigations. Lord et al. (2003) developed similar, basic apparatus to record body displacement directly at waist level. This sway meter comprises a rod with a vertically mounted pen at its end, attached to participants by a belt, and extending posteriorly (Lord et al., 2003). However, sway measures from this device (Lord et al., 2003) were reported to be strongly associated with CoP measures extracted from force platforms (Sherrington, 2000).

Table 2.1: Design, construction and materials of textured surfaces and foot insoles explored in current literature

Author	Surface/Insole Material	Indentations	Stimulated Foot Regions
Watanabe and Okubo (1981)	Transparent plate overlaid with evenly spaced shot gun balls	Shotgun balls; 2mm diameter, placed 10, 15 and 20mm apart, protruding approximately 1mm from the surface of the plate.	Full plantar aspect of the foot
Nurse et al. (2005)	3mm thick ethylene vinyl acetate foam, cut from a commercially available sandal	Semi-circular mounds with centre-to-centre distances of 8mm	Full plantar aspect of the foot
Corbin et al. (2007)	Plastic floor matting, from hardware store	Small rounded plastic ‘nubs’, raised approximately 0.25cm off the plastic surface	Full plantar aspect of the foot
Wilson et al. (2008)	Sheets of 3mm thick EVA (Shore Value A50)	*Dimpled texture*: 1mm raised circles, 3mm diameter, 5mm centre-to-centre distance *Grid texture*: 1mm raised square pyramid shapes with 2.5mm side length, 2.5mm peak-to-peak distance	Full plantar aspect of the foot
Palluel et al. (2008)	Semi-rigid PVC	*Spikes under the medial arch of the foot:* 10mm height, 5mm diameter with a density of 2 spikes/cm^2*. Spikes under remainder of the foot:* 5mm height, 3mm diameter with a density of 4 spikes/cm^2	Full plantar aspect of the foot
Kelleher et al. (2010)	Grade P80 Wet and Dry Sandpaper	Course sandpaper adhered to leather insole	Full plantar aspect of the foot

Textured foot insoles, cut from large commercial sheets of flooring (Corbin et al., 2007), shoe soling material (Wilson et al., 2008), and footwear comprising a spike insole (Palluel et al., 2008) have been investigated regarding their influence on postural stability during quiet standing (Corbin et al., 2007, Palluel et al., 2008, Wilson et al., 2008) and gait (Palluel et al., 2008, Wilson et al., 2008). Corbin et al. (2007) investigated the effect of textured foot insoles, comprising a geometric pattern of small rounded nubs (Table 2.1), on balance performance in 33 healthy young adults (17 female, age mean 27.4, SD 9.1 years) during double and single leg standing, under different visual conditions. During bilateral standing without textured insoles, participants' CoP area significantly increased by a mean difference of 0.7cm with eyes closed, compared to standing with eyes open (p=0.008). With textured insoles, average CoP area differed by only 0.1cm between visual conditions. This was not statistically significant (p=0.60). Similar findings were reported for mean CoP velocity. During bilateral standing without textured insoles, CoP velocity increased by 0.2cm s^{-1} with eyes closed, compared to eyes open (p<0.001). The difference in CoP velocity between visual conditions when wearing textured insoles was 0.1cm s^{-1} and not statistically significant (p=0.06). It appears that a mean difference of 0.1cm s^{-1} in CoP velocity is sufficient to distinguish between non-significant and significant findings. It is possible this would not be considered clinically important, yet the statistical analysis reports otherwise. This emphasises the limitation in solely relying upon *p* values for interpreting clinically relevant findings. However, the study by Corbin et al. (2007) shows that declines in balance performance, normally seen with removal of vision, were absent in measures of CoP area and velocity, when participants wore the textured insoles. These authors concluded that their textured insoles effectively compensated for a reduction in balance when vision was removed by closing the eyes (Corbin et al., 2007). The clinical significance of the textured effect on CoP measurements with eyes closed, may be that it is of similar magnitude to the effect of opening the eyes (Corbin et al., 2007). This is a clinically important finding as the textured insoles investigated by Corbin et al. (2007) may have the capacity to alter the position of the CoP and also 'buy' individuals more time to execute corrective postural responses, when visual sensory information is limited.

As all participants wore thin cotton socks during test procedures (Corbin et al., 2007), it is possible optimum plantar stimulation of mechanoreceptors was not achieved. There appears to be a trade-off between maximising plantar tactile stimulation and replicating normal conditions. Western adults commonly wear hosiery within footwear, therefore Corbin et al. (2007) chose test conditions which best represented daily life. However, it remains unknown whether the magnitude of the textured effect would have been greater with participants standing barefoot. What that research (Corbin et al., 2007) does indicate is that textured insoles have the capacity to alter balance performance, even when textured indentations are not in direct contact with the plantar surface of the foot.

The textured insoles investigated by Corbin et al. (2007) were not compared to a control insole, but rather a shoe only condition. Confounding factors including differences between insole hardness, in-shoe volume and foot pressure, could have influenced sway measures. Furthermore, as Corbin et al. (2007) measured balance over a short duration, 10 seconds, it is possible their significant finding may represent highly variable postural corrections (Le Clair and Riach, 1996), rather than a true effect of the texture. This would be consistent with evidence from Palluel et al. (2008) who concluded textured insoles have no immediate effect on postural sway during quiet standing with eyes closed.

Wilson et al. (2008) investigated the long-term effects of textured foot insoles on sway parameters during unperturbed quiet standing (AP and ML range) and gait trials (base of support) in 40 healthy middle-aged females (age mean 51.1, SD 5.8 years). The insoles were cut from three different shoe soling materials; a dimpled insole comprising convex hemispheres, a grid insole consisting of a raised pyramidal pattern, and a plain, smooth, flat-surfaced insole (Table 2.1) (Wilson et al., 2008). A control group wore commercially-available shoes fitted with a standard insole, yet there was no standardisation of insole hardness between the control and intervention conditions. With the textured foot insoles being more rigid, this could affect shoe and insole flexibility, and thereafter foot motion, particularly when walking. Increasing shoe midsole hardness has shown to increase ML movement of the CoM during standing after unexpected gait termination (Perry et al., 2007). However, Wilson et al. (2008) reported that on average, their textured insoles showed no statistically significant short-term effects at baseline on AP ($p=0.702$) or ML ($p=0.699$) sway or base of support ($p=0.699$) parameters. This was also the case after wearing the textured insoles for 4 weeks (AP sway: $p=0.091$, ML sway: $p=0.605$, Base of Support: $p=0.238$).

The authors concluded that their textured insoles showed no detrimental effects on postural stability in healthy middle-aged females (Wilson et al., 2008). However, with limited sway and gait data presented, in particular the absence of mean values, it is difficult to determine whether their non-significant changes in balance measures could be considered clinically significant or therapeutically beneficial. Box plots showing median changes in sway between foot insole conditions indicate that the dimple insole may have the greatest potential to enhance balance performance. During quiet standing with eyes open, the dimple insole appeared to decrease median ML sway, relative to control. Upon closing the eyes, the dimple insole also maintained a similar level of ML sway to the control group, thus had no clinically detrimental effect on balance. This is interpreted on the basis that increased sway indicates a decrease in postural stability. A clinically undesirable trend for greater AP sway was seen when wearing plain, grid or dimple insoles, relative to control, and irrespective of visual condition: this did not reach statistical significance. Also, during gait trials, participants wearing any one of these three foot insoles appeared to walk with a narrower base of support than the control group (Wilson et al., 2008). The clinical relevance of this finding is that a reduction in the base of support during walking indicates that the CoM has to be more tightly controlled within a smaller area, thus increasing the balance challenge and likelihood of experiencing instability.

A number of factors may have influenced the textured effect in the study by Wilson et al. (2008) including; limited number of static and dynamic balance measures analysed, unchallenging nature of the balance tasks, use of standardised footwear, and poor compliance with long-term insole wear.

Both Corbin et al. (2007) and Wilson et al. (2008) investigated textured insoles with protrusions extending the full length of the foot insole, providing continuous tactile stimulation of fore-, mid- and rear-foot regions. Thus, the construction of these foot insoles (Corbin et al., 2007, Wilson et al., 2008) fits with the definition of total and even stimulation (Watanabe and Okubo, 1981). Corbin et al. (2007) did not provide details regarding the spacing of protrusions specific to their textured insole. The dimple insole investigated by Wilson et al. (2008), appeared to show greatest potential for improving balance performance, comprised protrusions spaced 5mm apart. The evidence from Watanabe and Okubo (1981) and Wilson et al. (2008) suggest there is an optimal range within which protrusions should be spaced, in order for textured surfaces or insoles to significantly affect balance. It remains unknown whether intermittent, selective stimulation of peripheral regions or complete stimulation of the full plantar aspect of the foot leads to greater improvements in balance performance.

Palluel et al. (2008) explored the effect of wearing spike insoles within sandals when standing quietly with eyes closed over 5 minutes. Relative to previous studies (Corbin et al., 2007, Watanabe and Okubo, 1981, Wilson et al., 2008) this work (Palluel et al., 2008) presented the potential to explore habituation to texture over a longer duration. The term ‘spike insoles’, referred to a commercially available design of footwear used for pool activities, comprising an insole covered with semi-rigid PVC spikes (Table 2.1). In comparison to a control condition, spike insoles caused some significant reductions in sway amplitude and speed in young and older participants (Palluel et al., 2008). This change in balance is clinically beneficial as the CoP is moving at a slower speed over a shorter distance, thus upright balance is more easily maintained, reducing the demands placed on the postural control system.

During quiet standing, spike insoles significantly reduced ML root mean square ($p=0.036$) in 8 out of 19 young adults (9 females, age mean 25.9, range 21-32 years), compared to control. When walking with spike insoles, significant improvements were reported, in a proportion of the total 19 young participants investigated, for: CoP surface area ($p=0.001$; $n=15$), AP root mean square ($p=0.027$; $n=15$) and ML root mean square ($p=0.036$; $n=13$) (Palluel et al., 2008).

However, spike insoles appeared to have slightly greater therapeutic effects in some of the 19 healthy older people investigated (11 females, age mean 69.0, range 62-80 years). Significant reductions in sway measures, exclusive to quiet standing tests included: CoP surface area ($p=0.001$; $n=14$), CoP speed ($p=0.005$; $n=10$), AP ($p=0.003$; $n=11$) and ML root mean square ($p=0.036$, $n=12$). The significant effects of spike insoles on CoP speed and ML root mean square in older people were also observed during walking trials (Palluel et al., 2008).

As the authors (Palluel et al., 2008) do not present absolute mean sway data for the spike or control conditions, it is difficult to discern the magnitude of the clinical effect of their textured insoles. More so, one cannot evaluate whether the construction or geometry of the spike insole (Palluel et al., 2008) is of greater therapeutic benefit than those explored in previous studies (Corbin et al., 2007, Watanabe and Okubo, 1981, Wilson et al., 2008).

Palluel et al. (2008) concluded that immediate exposure, defined as the first 32 seconds after putting the spike insoles on, brought about no changes in postural sway in young or older participants. However, evidence suggests postural sway measures significantly differ over 10, 20 and 30 second test periods (Le Clair and Riach, 1996) and over the course of a 60 second quiet standing test (Samson and Crowe, 1996). Therefore, it is possible 32 seconds was too long a duration to capture the true immediate textured effect on postural responses.

Current literature indicates textured insoles do not affect balance performance in healthy young and middle-aged adults when standing with eyes open (Corbin et al., 2007, Wilson et al., 2008). Palluel et al. (2008) only performed tests of quiet standing with eyes closed, based on previous evidence concluding vision to be the predominant source of sensory information for balance control in older adults (Meyer et al., 2004, Perrin et al., 1997). Therefore, the effects of texture may only become apparent when balance is sufficiently challenged, through manipulation of sensory information or application of an external perturbation. However, the findings from Corbin et al. (2007) may dispute this suggestion. These authors observed that increasing the balance challenge from double- to single-limb standing, their textured insoles had no effect on postural sway measures (Corbin et al., 2007). This evidence may suggest that when balance is overly challenged, any textured effects are overridden. This could point to a possible upper and lower limit of the effects of texture, depending on how much balance is affected in the first place.

Significant effects of texture on postural sway measures in healthy adults have been unanimously observed during quiet bilateral standing with eyes closed (Corbin et al., 2007, Palluel et al., 2008, Watanabe and Okubo, 1981). This seems to indicate that enhancing afferent sensory information to the plantar surface of the feet can act as a surrogate source of input for controlling postural sway, when visual information is suppressed (Corbin et al., 2007). These findings, in healthy adults, may support the potential for texture to have greater effects on postural control in people with pathological or age-related balance deficits.

2.9: Texture and Muscle Activity

To date, only two studies have investigated the influence of texture on lower limb muscle activity (Kelleher et al., 2010, Nurse et al., 2005).

Nurse et al. (2005) explored the effect of textured foot insoles comprising semi-circular mounds, on gait patterns (Table 2.1). 15 healthy young adults (3 females, age mean 24.7, SD 2.9 years) conducted repeated gait trials at a standardised speed whilst wearing the textured foot insoles. Significant decreases were observed in overall EMG intensity of soleus and tibialis anterior, for the entire stance phase with the textured foot insole, relative to a control condition. Nurse et al. (2005) reported a 13% decrease in soleus activity during the propulsive phase of stance, and a 13% decrease in tibialis anterior activity immediately following heel strike. Data from each lower limb muscle, during walking with textured foot insoles, were normalised to each participant's overall EMG intensity for the control condition.

Analysing mean EMG amplitude within specific frequency bands, concluded that reductions in soleus and tibialis anterior intensity occurred within the low frequency spectrum, with only soleus reaching the level of significance ($p<0.05$) (Nurse et al., 2005). On average, no other significant differences were observed for any other muscle or time interval. This may suggest the sensory effects of textured foot insoles are primarily influential on slower motor units (Wakeling et al., 2002), given their slower conduction velocity would lower the frequency content of the EMG signal (Solomonow et al., 1990). However, during steady, level-ground walking it would be expected that slow twitch fatigue-resistant muscle fibres would be recruited.

No significant differences were reported for mean thigh muscle activity between texture and control conditions (Nurse et al., 2005). This suggests textured foot insoles may only have the capacity to reduce the demand placed on primary foot invertors (tibialis anterior and soleus), or distal muscles located close to the site where tactile stimulation is applied. However, it is possible textured insoles could alter activity in proximal lower limb muscles, not investigated in that study (Nurse et al., 2005).

Kelleher et al. (2010) explored the effect of textured foot insoles on lower limb EMG activity during walking, in 14 middle-aged adults with Multiple Sclerosis (6 females, age mean 41.8, SD 7.3 years). The textured insoles investigated in that study, were constructed from sandpaper, fixed to fine leather insoles (Table 2.1). Participants were barefoot so that the course sandpaper was in direct contact with the plantar surface of the feet, within shoes.

Mean EMG activity (as a percentage of peak amplitude) was collected for tibialis anterior, soleus, medial and lateral gastrocnemius, over three phases of the gait cycle. When wearing textured insoles, participants showed significantly greater mean (SD) EMG activity in lateral gastrocnemius from heel-strike to peak vertical ground reaction force (27.9 ± 13.4%) compared to no insole (25.9 ± 13.0%). Increased lower limb EMG activity during the stance phase of gait may be coupled with observed increases in knee flexion during this same period, when wearing textured insoles (Kelleher et al., 2010). Enhanced plantar tactile stimulation may trigger such kinetic and kinematic responses as contact between the plantar surface of the foot and textured insole will be greatest when the heel, then forefoot, impact the ground. However, in participants with Multiple Sclerosis, consideration must also be given to the presence of lower limb spasticity and potential triggers, such as tactile stimulation, which may exacerbate levels of muscle tone.

During the swing phase of gait, walking with textured insoles brought about significant increases in medial gastrocnemious activity (16.4 ± 8.9%), compared to control conditions (15.2 ± 7.8%) (Kelleher et al., 2010). This evidence indicates that providing enhanced tactile stimulation to the feet can alter lower limb muscle activity during both stance and swing phases of the gait cycle (Kelleher et al. 2010). This points to two possible interpretations; firstly, the foot does not need to be in contact with the ground for the textured effect to take place, or alternatively, textured insoles may cause a muscular response in the contralateral limb (Tax et al. 1995). Although participants with Multiple Sclerosis were reported to show impaired plantar sensation, particularly at the medial aspect of the foot, textured insoles still appeared to significantly affect kinetic and kinematic parameters (Kelleher et al., 2010).

The findings from Nurse et al. (2005) and Kelleher et al. (2010) suggest textured insoles can significantly alter EMG activity in muscles of the lower leg when walking, in both healthy and pathological groups.

2.10: Texture and Underlying Mechanisms

Enhancing plantar sensory information at the feet, using texture, appears to influence postural control and muscle activation patterns in adults (Corbin et al., 2007, Kelleher et al., 2010, Nurse et al., 2005, Palluel et al., 2008, Watanabe and

Okubo, 1981, Wilson et al., 2008). However, the mechanism by which a textured surface or foot insole may initiate such changes remains unknown.

Textured surfaces may provide sufficient tactile stimulation to alter the rate of discharge from mechanoreceptors or spatio-temporal firing patterns of populations of sensory afferents located in the feet (Nurse et al., 2005). Plantar mechanoreceptors are perceptive to stimuli such as stroking, indenting or stretching of the skin due to contact with a surface (McGlone and Reilly, 2010, Sherwood, 1997), and changes in the acceleration or location of foot pressure (Kavounoudias et al., 1998). Therefore tactile stimuli, detected by plantar cutaneous mechanoreceptors, can provide the CNS with vital information regarding the location of peak foot pressure, relative to alterations in upright body position (Kavounoudias et al., 1998).

Afferent nerves originating from mechanoreceptors located on the plantar surface (Kennedy and Inglis, 2002) or lateral border (Trulsson, 2001) of the foot are distributed to either slow- (Merkel discs and Ruffini endings) or fast-adapting (Meissner or Pacinian corpuscles) units (Vedel and Roll, 1982). The relevance of this distribution to the current thesis is that slow-adapting units respond to maintained skin indentation with regular or irregular discharges. Fast-adapting units show burst responses to stimuli, only at the point of application or removal from the foot (Kennedy and Inglis, 2002). Therefore, prolonged standing on a textured surface would most likely alter discharge patterns from slow adapting mechanoreceptors. As the feet remain in contact with the supporting surface during bilateral stance, slow-adapting mechanoreceptors may have a major role in detecting changes to foot contact patterns (Trulsson, 2001) and pressure distribution relative to CoP movement. However, 70% of plantar mechanoreceptors in healthy young adults are reported to be fast-adapting (Kennedy and Inglis, 2002). Similarly, over half the mechanoreceptors identified on the lateral border of the foot were classified as fast-adapting (Trulsson, 2001).

The plantar surface of the foot has been described as a ‘dynamometric map’ for human balance control following investigations exploring selective stimulation of mechanoreceptors (Kavounoudias et al., 1998). Applying vibratory probes to selected plantar foot regions in healthy adults, during quiet standing, has shown that the direction of CoP movement is specific to the foot area stimulated (Kavounoudias et al., 1998). Vibration was applied at the forefoot and/or rearfoot of one or both plantar surfaces. Stimulating one region under one foot caused the body to tilt in the opposite direction to the site of vibration. Stimulating two regions of either the same or different plantar surfaces has been shown to induce body tilting in a direction contralateral to the site of stimulation (Kavounoudias et al., 1998). It is possible that applying vibratory stimulus at a specific foot region could mimic local increases in foot pressure, as experienced when the body is actually tilted towards this region. Thus, the body maintains upright balance by tilting in the opposite direction (Kavounoudias et al., 1998). However, it remains unknown whether the 'dynamometric map' of the plantar surface of the foot is the same in healthy individuals as those presenting pathologies, or in young and older adults.

This evidence (Kavounoudias et al., 1998) and other studies (Galica et al., 2009, Hijmans et al., 2007b, Hijmans et al., 2008, Priplata et al., 2003) suggest applying mechanical or electrical noise to plantar mechanoreceptors may have the capacity to influence postural responses. This phenomenon, known as stochastic resonance, suggests applying sub-sensory noise can enhance detection, transmission and flow of weak signals, thereafter maximising sensorimotor function (Gravelle et al., 2002, Priplata et al., 2003). However, it is unlikely that textured surfaces work on this same principle as they provide neither mechanical nor electrical stimuli. A systematic review by Hijmans et al. (2007a) acknowledges that vibratory stimuli may affect intrinsic foot proprioceptors, in addition to cutaneous mechanoreceptors. However, tactile stimulation from vibratory probes, placed directly against the skin, may have been sufficient to alter postural sway, irrespective of the vibratory component (Priplata et al., 2003). This would support findings from previous studies investigating textured insoles without vibratory components (Corbin et al., 2007, Palluel et al., 2008).

Previous studies investigating textured surfaces or foot insoles provide data for either balance performance or lower limb EMG activity: this evidence is limited in older adults. Exploring the effect of texture on balance and muscle activity may provide important information relating to the effect of texture on sensorimotor function. It remains unclear whether changes in balance are accompanied by underlying alterations in muscle activity. This thesis will present novel data for the effect of textured surfaces on postural sway and lower limb EMG activity in healthy young and older adults.

2.11: Summary

Optimal sensorimotor function is vital for balance control: this can be disrupted through acute injury, chronic pathology and increasing age. Age-related decline in physiological systems leads to significant differences in postural control mechanisms between young and older adults. Somatosensory information plays a major role in maintaining upright balance in healthy adults. A number of studies have explored the potential of footwear interventions to improve balance by providing enhanced plantar tactile stimulation to cutaneous mechanoreceptors. There is some evidence to suggest that textured insoles and surfaces have the capacity to alter balance performance in healthy young adults, with research into older people being limited. Other studies suggest textured foot insoles can significantly alter lower limb muscle activation patterns during walking. The mechanism by which texture alters measures of postural sway and muscle activity remains unknown. No current research has explored the effects of textured surfaces on postural sway and lower limb muscle activity in healthy young and older adults. It remains unknown whether alterations in lower limb muscle activity is an underlying mechanism to changes in postural sway. Previous studies have explored the effect of only one design of textured intervention compared to control, thus it is unclear whether an optimal geometric textured pattern exists for maximising balance performance.

2.12: Aims of the Thesis

2.12.1: Research question

Do textured surfaces affect postural sway and lower limb muscle activity during quiet standing in healthy young and older adults?

2.12.2: Specific objectives

The current thesis explores whether textured surfaces have a role to play in enhancing short-term balance performance in healthy young and older adults by way of altering the following postural sway measures: AP and ML standard deviation, AP and ML range and CoP velocity.

Two different textured surfaces, and one smooth surface as control, are explored to determine the importance of the geometric design of a textured surface in altering balance performance. The two textured surfaces are compared against the control condition to determine whether there is a specific textured pattern which has the capacity to enhance balance performance relative to a 'no texture' condition. This comparison reflects a clinically relevant scenario, that being, a comparison of balance performance at baseline versus intervention.

The current thesis explores the effect of textured surfaces on postural sway and lower limb muscle activity simultaneously during quiet standing under two different visual conditions: eyes open and eyes closed. This is to determine whether textured surfaces provide a source of plantar tactile stimulation which has the capacity to enhance balance when visual information is available, or compensate for a loss of balance due the removal of visual information when the eyes are closed.

This thesis will explore the effect of textured surfaces on postural sway and lower limb muscle activity over three different time intervals: the first 10, latter 20 and overall 30 seconds of quiet standing. This is to determine whether a textured intervention shows to be clinically effective in enhancing balance during or following re-stabilisation of the body, after standing up from a seated position.

The current thesis also explores whether textured surfaces alter lower limb muscle activation as an underlying mechanism to changes in postural sway.

The current thesis explores the influence of foot posture as a factor in the effect of textured surfaces on postural sway and lower limb muscle activity. This is to determine whether the contact area between the plantar surface of the foot and the textured surface in individuals with normal, high-arched and low-arched foot posture, influences the magnitude of the textured effect.

The current thesis will provide exploratory findings indicating the potential for texture to be incorporated into possible clinical interventions designed to enhance balance, such as foot insoles.

The current thesis comprises two separate studies:

Phase 1: The effect of textured surfaces on postural sway and lower limb muscle activity during quiet standing in healthy young adults

Phase 2: The effect of textured surfaces on postural sway and lower limb muscle activity during quiet standing in healthy older adults

2.12.3: Hypotheses

The current thesis has the following null hypotheses for both healthy young and older adults:

Postural Sway:

- Texture 1 will not alter postural sway parameters during randomised conditions of quiet standing with eyes open or eyes closed over the first 10, latter 20, or overall 30 seconds, relative to the control surface.
- Texture 2 will not alter postural sway parameters during randomised conditions of quiet standing with eyes open or eyes closed over the first 10, latter 20, or overall 30 seconds, relative to the control surface.

Lower limb EMG activity:

- Texture 1 will not alter lower limb EMG activity during randomised conditions of quiet standing with eyes open or eyes closed over the first 10, latter 20, or overall 30 seconds, relative to the control surface.
- Texture 2 will not alter lower limb EMG activity during randomised conditions of quiet standing with eyes open or eyes closed over the first 10, latter 20, or overall 30 seconds, relative to the control surface.

Foot Posture:

- The effect of Texture 1 on postural sway parameters and lower limb muscle activity during randomised conditions of quiet standing with eyes open or eyes closed over the first 10, latter 20, or overall 30 seconds, relative to the control surface, will not differ between normal, high-arched or low-arched foot postures.
- The effect of Texture 2 on postural sway parameters and lower limb muscle activity during randomised conditions of quiet standing with eyes open or eyes closed over the first 10, latter 20, or overall 30 seconds, relative to the control surface, will not differ between normal, high-arched or low-arched foot postures.

CHAPTER 3: REVIEW OF MEASUREMENT TECHNIQUES

3.1: Introduction

This chapter will critically appraise and evaluate techniques used in the current study, to test the experimental hypotheses. Justification for use of the force platform and electromyography instrumentation will be presented. The chapter will conclude discussing participant screening assessments and the choice of textured surfaces.

3.2: Postural Stability

There are three main approaches to the measurement of balance performance (Horak, 1997). A functional approach focuses on an individual's ability to perform tasks relevant to daily life (Shumway-Cook and Woollacott, 1995). A systems approach sets out to determine underlying pathological causes of balance impairment. Thirdly, quantitative posturography uses biomechanical instrumentation to measure postural sway, providing information about small body adjustments occurring during balance tests (Horak, 1997).

In the current thesis, balance performance during quiet standing on two different textured surfaces, and one smooth surface as control, will be assessed using static quantitative posturography. This approach can be justified by a number of reasons. Firstly, the effects of introducing a new interface, proposed to alter somatosensory input between the ground and feet, are unknown and must initially be explored during unchallenging balance tasks, such as quiet standing. Similar methodology has been used in previous research, concluding that quiet standing is a substantial test for determining the effects of sensory manipulation on balance performance (Corbin et al., 2007, Laughton et al., 2003, Raymakers et al., 2005). Only once this information has been established would it be safe to explore the effect of texture during more demanding functional activities. Secondly, as the current thesis investigates the effect of textured surfaces on postural stability in healthy young and older people, with no known underlying pathologies or balance impairments, a systems approach to balance measurement is unnecessary.

3.2.1: Introduction to the Kistler force platform

Postural sway data is commonly collected from a force platform. The current study used a Kistler force platform, which comprises triaxial piezoelectric force transducers embedded within each of its four corners. When standing upright, the feet remain continually in contact with the supporting surface. A three-dimensional force vector, known as the ground reaction force, acts on each foot and defines the position of the average pressure point (Richards and Thewlis, 2008). The location of these ground reaction force vectors on the floor is termed the CoP. Force platforms are used to quantify the location of the CoP (Richards and Thewlis, 2008). As participants load the force platform, the piezoelectric crystals become electrically charged. This charge is proportional to the force acting upon the crystals, from which various components of the ground reaction force can be captured.

During bilateral standing, there are two ground reaction force vectors, one under each foot. Each ground reaction force can be measured individually using two separate force platforms (one for each foot), with such methodology allowing for observation of asymmetrical loading (Rougier, 2008). However, in the current thesis, one force platform will be used to calculate the position and movement of the net CoP. Net CoP refers to the summation of the two ground reaction force vectors acting under the feet, thus showing the position of the combined forces (Richards and Thewlis, 2008). The current thesis did not aim to analyse loading patterns, but rather generate a clinically relevant summary of postural stability on a global level.

Force platforms are also used for measuring dynamic postural stability, whereby the force platform can be sway referenced or tilted to provide external perturbation, or used in combination with moving visual surrounds to assess the effects of sensory manipulation. The current thesis does not aim to assess the effect of textured surfaces on postural stability during dynamic conditions. Furthermore, sensory manipulation will be achieved by way of introducing different textured surfaces and by closing the eyes. Therefore, in the current thesis, postural sway data will be collected from one static Kistler force platform (Model 9286AA, Kistler, Alton, UK).

3.2.2: Kistler force platform sampling rate

When postural sway data is sampled at too low a frequency, this can lead to substantial loss of information. Multiple sway parameters sampled at 50Hz have shown to be considerably greater in terms of their length or speed, compared to sampling the same data at 10Hz (Raymakers et al., 2005). In the current thesis postural sway data will be sampled at a rate of 1000Hz (Bergland and Wyller, 2004). This procedure will minimise loss of important sway data and also ensure consistency with the sampling rate of EMG data.

3.2.3: Centre of pressure as a representation of postural stability

The current thesis reports CoP movement resulting from enhanced plantar tactile stimulation. Winter (1995) modelled the body on an 'inverted pendulum', considering the ankle to be the only moveable joint, acting as a fulcrum around which the CoM drifts. The CoM is considered to be an imaginary point at which total body mass is concentrated (Lafond et al., 2004). The CoP always oscillates beyond the CoM, in order to maintain the latter in a relatively central position (Winter et al., 1998). This relationship indicates the CoP controls the CoM, and changes in CoP position will reflect alterations in CoM location (Winter, 2005).

There is general agreement that increased CoP movement, or postural sway, represents a decline in postural stability. As highlighted in chapter 2 of this thesis, such claims are based on evidence that CoP movement can increase in the presence of injury (Bonfim et al., 2008), chronic pathology (Hassan et al., 2001, Hinman et al., 2002, Mann et al., 2009, McKeon and Hertel, 2008, Mientjes and Frank, 1999, Tjon A Hen et al., 2000), with increasing age (Colledge et al., 1994, Maki et al., 1990, Maki et al., 1994), and is associated with a risk of falling (Maki et al., 1994). By reducing the amplitude and increasing the frequency of CoP movement, this may assist adults to retain their CoM within the limits of stability, when upright balance is threatened (Adkin et al., 2000). In contrast, increased CoP movement during quiet standing, may provide a mechanism to augment proprioceptive sensory input from the lower limbs to the CNS (Schieppati et al., 1994), and enhance balance (Van Emmerik and Van Wegen, 2002).

Changes in postural sway should be interpreted with caution, giving consideration to the context in which balance performance is assessed. Tests of quiet standing are considered sufficiently robust to detect changes in balance control, yet may lack functional relevance as this task rarely occurs in isolation, but rather integrated within activities of daily living (Van Emmerik and Van Wegen, 2002).

3.3: Postural Sway Parameters

CoP movement, commonly referred to as postural sway, is defined as the small forward and back (AP) and side to side (ML) shifts in body position, when standing upright (Shumway-Cook and Woollacott, 1995). The magnitude of CoP movement provides a measure of balance ability (Hasselkus and Shambes, 1975) and the steadiness of an individual. In the current thesis, measures of CoP amplitude and velocity will be reported, generating evidence for the effect of textured surfaces on sway variability, amplitude and velocity during quiet standing.

3.3.1: AP and ML range

The amplitude of CoP movement during quiet standing can be quantified using AP and ML range; defined as the distance, in mm (from point to point), between the extremes of sway in each direction. Therefore, sway range quantifies the distance between the minimum and maximum CoP positions (Hertel and Olmsted-Kramer, 2007), thus using only two data points from an entire data set (Palmieri et al., 2002). It is worth noting that the minimum and maximum CoP positions will most likely show variance between trials and may not represent all changes occurring in the data set for one quiet standing trial. However, AP and ML range may importantly, be related to the functional base of support (Nigg et al., 2006). Large AP and ML sway range may only become clinically problematic when the CoP moves beyond the limits of stability.

Previous studies have chosen to measure the magnitude of CoP movement, in order to quantify the effects of increasing age, gender (Amiridis et al., 2003, Bryant et al., 2005, Era et al., 1997, Panzer et al., 1995), and the application of biofeedback (Vuillerme et al. 2008) on postural stability during quiet standing.

Sway parameters measuring the amplitude of CoP movement, such as AP and ML range, may relate to the effectiveness of the postural control system (Maki et al., 1990), represent neuromuscular responses to postural imbalances (Winter, 1990), and can predict future risk of falling (Maki et al., 1994, Stel et al., 2003). AP and ML range are clinically useful sway parameters, as it is only when the CoM moves beyond the limits of stability that the body may fall over, should a postural control mechanism fail.

3.3.2: AP and ML standard deviation

AP and ML standard deviation (SD) is a time-independent variable, quantifying the degree of dispersion in CoP position, about its mean location. The SD is the square root of the variance, measuring the spread of the data about the mean CoP position. Therefore the SD is sometimes referred to as the root mean square amplitude (Palmieri et al., 2002). The SD of CoP movement is a standard measure used in the field of balance performance to describe participants' postural behaviour during quiet standing (Vuillerme et al., 2008). Vuillerme et al. (2008) indicate that calculating the SD of CoP displacements in both AP and ML directions, generates a measure of spatial variability of the CoP about its mean position. Within current literature, it remains unclear whether more or less sway variability is beneficial to postural control (Haddad et al., 2008, Nagy et al., 2007, Van Emmerik and Van Wegen, 2002). However, parameters such as AP and ML SD may be inherently related to functionally relevant aspects of postural stability including; exploratory postural behaviour and the number of solutions available to make postural corrections when balance is threatened.

Pinsault and Vuillerme (2009) explored the test re-test reliability of CoP surface area: this parameter calculates the surface area covered by CoP with a 90% CI (in mm^2). It has been suggested that this parameter is similar to CoP SD, generating information relating to CoP spatial variability (Pinsault and Vuillerme, 2009). Variance of the position of the CoP in both AP and ML axes (in mm^2), was also calculated, but presents a different aspect of sway variability relative to CoP SD. However, these parameters were chosen given they are easy to use, computationally undemanding, and commonly adopted in clinical practice to determine an individual's postural control capacities (Pinsault and Vuillerme, 2009). Other studies have reported AP and ML sway variability increases the longer the test duration, ranging from 10 to 60 seconds, in healthy young adults (Le Clair and Riach, 1996). In the short-term, the CoP may circulate round a single target within the base of support, thereafter extending to multiple targets thus increasing variability of CoP movement (Le Clair and Riach, 1996).

The SD of CoP movement has shown to be a reliable parameter for assessing postural control (Le Clair and Riach, 1996). Previous studies have shown the SD of CoP movement to be sensitive in detecting alterations in balance performance due to age (Amiridis et al., 2003, Carpenter et al., 2006, Hsiao-Wecksler et al., 2003, Panzer et al., 1995) and the application of electrotactile biofeedback (Vuillerme et al., 2008).

Panzer et al. (1995) used the magnitude and variability (calculated using the SD) of CoP movement to explain observed age-related changes in postural control, as these sway parameters were considered to be sensitive measures of quiet standing ability. Similarly, Vuillerme et al. (2008) used the SD and range of CoP movement to show the effect of placing an electrotactile biofeedback system on the tongue, on quiet standing balance performance.

In a review by Palmieri et al. (2002), which considers a small number of CoP parameters used for assessing postural control, these authors indicate that variation in AP and ML SD (or root mean square amplitude) can be used to identify changes in postural control. However, it remains unknown what changes are occurring within the postural control system, leading to observed alterations in AP and ML SD.

3.3.3: CoP velocity

CoP velocity is a time-dependent sway parameter, providing important information about the rate at which the CoP moves within, and about, the base of support and could be a major determinant in balance maintenance (Van Emmerik and Van Wegen, 2002). CoP velocity is defined as the total distance covered by the CoP, divided by the sampling period (Simoneau et al., 2008). It is considered the most consistent stability parameter and most sensitive for detecting changes in balance, influenced by age and availability of visual information (Prieto et al., 1996, Raymakers et al., 2005). CoP velocity is suggested to be proportionate to the level of activity required to regulate postural stability (Maki et al., 1990).

CoP velocity is considered to have better discrimination to changes in postural stability, compared to AP and ML SD and sway range. Changes in balance performance in older people have shown to be more sensitive to measures of CoP velocity (Prieto et al., 1996), than variables quantifying the distance of CoP movement (Prieto et al., 1996, Raymakers et al., 2005). The greater sensitivity of CoP velocity to age-related changes in balance could be attributed to its time-dependency or multidirectional measurement, as net CoP velocity represents combined sway data from both AP and ML co-ordinates. This could correspond to age-related changes in compensatory postural control mechanisms. Older people may have more complex postural movement strategies, which may be more sensitive to sway parameters based on multidirectional measures (Vaillancourt and Newell, 2002). Additionally, during quiet standing, sensory inputs provide more precise information about the velocity at which the body is moving, rather than its position or acceleration (Kiemel et al., 2002). Body movement velocity is considered the most accurate form of sensory information for stabilising upright posture (Jeka et al., 2004).

Research has shown that postural sway is a dynamic process and its spectral characteristics can change over time (Loughlin and Redfern, 2001). Therefore, time-dependent sway variables such as CoP velocity or more complex analytical methods including spectral frequency (Loughlin and Redfern, 2001) may be more effective in detecting subtle changes in postural stability between conditions, uncovering time-varying amplitudes of sway data. Williams et al. (1997) suggest spectral frequency analysis may have greater capacity to detect age-related deterioration in postural control systems, beyond traditional sway measures, particularly as older adults have shown a characteristic pattern of slow spontaneous lateral shifting of body weight during quiet standing (Williams et al., 1997). However, the current thesis does not set out to explore the long-term effects of textured surfaces on age-related changes in balance performance. In agreement with previous methods of data collection and analysis, traditional sway parameters are considered sufficiently sensitive and robust to explore balance performance.

Therefore, in the current thesis, AP and ML sway range will provide time-independent, unilateral measures of the amplitude of CoP movement, relative to the minimum and maximum location of the CoP. AP and ML SD provide an insight of sway amplitude variability. CoP velocity is a consistent sway parameter which considers sampling duration and may be more sensitive to changes in balance performance between textured conditions, particularly in older participants.

3.4: Factors Affecting Force Platform Data

3.4.1: Foot positioning

Altering foot position, and therefore the base of support, during repeated measures of bilateral quiet standing has been shown to significantly influence postural sway parameters (Day et al., 1993, Kirby et al., 1987). Greater stance width, defined as the heel-to-heel distance, can lead to more prominent reductions in CoP velocity and ML sway, compared to AP sway (Day et al., 1993). Bringing the feet closer together can increase ML sway, without affecting AP sway (Kirby et al., 1987), suggesting lateral sway variables may be more sensitive to changes in foot position. As ML sway is influenced by age (Raymakers et al., 2005) and a strong indicator of falls risk (Maki et al., 1994), this provides strong rationale for standardising foot positioning within the current thesis: particularly in older participants.

Foot orientation can also influence standing balance (Kirby et al., 1987). With the toes laterally oriented at 25º angles, the CoP has been shown to move only a short distance in AP and ML directions. In comparison, CoP movement was observed to be greatest with the toes medially oriented at 45º angles (Kirby et al., 1987). Thus, heel-to-heel distance may be a strong determinant of postural stability, especially if substantial loading occurs through the rearfoot.

Standardised foot position has been achieved using a variety of procedures including; pre-determined heel-to-heel distance (Benjuya et al., 2004, Nardone et al., 1995, Priplata et al., 2003) and toe angles (Benjuya et al., 2004, Priplata et al., 2003), foot templates (Laughton et al., 2003, Wilson et al., 2008), or tandem stance (Nichols et al., 1995). Such methods could lead to unnatural or uncomfortable stance positions, thereafter altering normal corrective balance responses. Self-selected, comfortable standing may avoid such constraints (Panzer et al., 1995, Stel et al., 2003), however, older adults prone to falling have adopted excessively wide stance widths when such methods were implemented (Maki et al., 1991).

This evidence justifies the need to standardise foot position in the current thesis. Therefore, foot position will be individually standardised for each participant to enhance the reliability of sway measures and minimise confounding alterations in lower limb joint kinematics.

3.4.2: Lower limb positioning

During quiet standing, the degree of flexion at the knees could have confounding effects on lateral balance control (Day et al., 1993). Full knee extension suppresses AP and ML joint movement, establishing a mechanical coupling mechanism between the ankle and hip joints, which no longer move independently (Day et al., 1993). Any movement occurring at the ankle joint translates proximally to the hip, leading to greater activation and stretch of surrounding hip musculature and proprioceptors (Day et al., 1993). With hip and ankle proprioceptors working together, this may enhance proprioceptive awareness and sensitivity to lateral motion originating at the ankle, contributing to improved overall ML balance control (Day et al., 1993).

In this thesis, following previous research methodologies, standing posture was standardised by asking participants to adopt their normal, comfortable quiet standing position (Panzer et al., 1995, Stel et al., 2003). This is justified as standardising lower limb position by predetermined criteria such as knee angle may not replicate participants' normal posture, or be possible in older adults with musculoskeletal conditions affecting the lower limbs, such as osteoarthritis and joint replacements.

3.4.3: Upper body positioning

In the current thesis, upper body positioning was not standardised during the transition from a seated to quiet standing posture. Acquisition of sway data did not commence until participants' trunk and lower limbs were perceived to be vertical. Only upon reaching this upright position, were participants instructed to stand with their arms hanging relaxed by their sides (Laughton et al., 2003, Mochizuki et al., 2006).

3.4.4: Head positioning

Previous research has shown that during quiet standing, postural stability in healthy older adults can deteriorate when the head is flexed or extended to 45º from vertical, (Buckley et al., 2005). Extending the head beyond vertical has been shown to: disrupt AP sway when standing with eyes closed (Paloski et al., 2006) and increase mean CoP velocity, AP and ML sway (Vuillerme and Rougier, 2005). This may occur as the utricular otollith organs become oriented beyond their working range, decreasing their sensitivity (Brandt et al., 1981). Alterations in head position during quiet standing can shift the CoM, thus disrupting postural alignment. This could affect ankle joint moments, demand counteractive lower limb muscle activity (Collins et al., 2003, Laughton et al., 2003), re-distribute plantar pressures and thus affect sensory input to plantar mechanoreceptors (Wu and Chiang, 1996). Therefore, in the current thesis, head position during quiet standing was standardised using a visual target, which was mounted on a board at each participant's eye level.

3.5: Electromyography

3.5.1: Origin of the electrical signal

Surface electromyography is a technique used to study muscle function, whereby myoelectrical activity during the excitation-contraction coupling process is recorded at the surface of the skin (Kamen and Caldwell, 1996). Therefore, EMG is the amplification of motor neuron activity between the spinal cord and muscles (Turker, 1993). The electrogram (EMG) provides visual output of muscle fibre activation. To understand the characteristics of the EMG signal, it is important to consider physiological processes contributing to muscle fibre activity.

A motor unit is the smallest contractile unit within skeletal muscle. It comprises an α-motorneuron, located in the spinal cord, and all the muscle fibres it innervates (Moritani et al., 2004). Motorneurons are activated by the CNS, following which an electrical impulse is propagated to each motor endplate (Kamen, 2004). Movement of ions at the postsynaptic membrane causes depolarisation, generating a muscle fibre action potential. This action potential propagates along the length of the muscle fibre. The sum of all muscle fibre action potentials, activated by one motorneuron, collectively generates a motor unit action potential, which is recorded at the surface of the skin (Kamen, 2004). Only motor unit action potentials close to the EMG electrodes will be detected (Buchthal and Rosenfalck, 1973). The amplitude of the EMG signal is largely attributed to the propagation velocity of the action potential (De Luca, 1979, Kamen and Caldwell, 1996) and the proximity of active muscle fibres, relative to the recording electrode (Buchthal and Rosenfalck, 1973). Action potential velocity will also govern how long the electrical signal is of sufficient strength to be detected at the surface of the skin (Kamen and Caldwell, 1996).

Repeated activation of motor units leads to a sustained muscle contraction. Depolarisation of muscle fibres, stimulated by one particular motor unit, overlaps with another. Thus, synchronised motor unit activity generates a complex pattern of electrical potentials (Loeb and Ghez, 2000), which contributes to the graphical presentation of the EMG signal. (Stegeman et al., 2000).

3.5.2: EMG amplitude

The amplitude of an EMG signal will change according to the intensity of a muscle contraction. During isometric contractions, increases in muscular force are linearly related to incremental increases in EMG amplitude (Kamen, 2004). Greater muscle force, and thus EMG amplitude, arise from a combination of increased motor unit firing rate and recruitment of additional motor units (Farina et al., 2004). However, the EMG-force relationship is not always linear. During dynamic muscle contractions the length of a muscle changes due to antagonistic concentric and eccentric activity, which will vary throughout the course of a contraction (Bigland and Lippold, 1954). The EMG-force relationship is affected by alterations in muscle tension, contraction velocity (Kamen, 2004), and variations in motor unit activity specific to contraction-type (Richards et al., 2008). Bigland and Lippold (1954) reported that faster contraction velocities greatly increase the EMG amplitude. Grabiner and Owings (2002) concluded that concentric contractions, whereby a muscle shortens, show a greater amplitude of EMG activity, compared to eccentric contractions.

The amplitude of an EMG signal can be affected by intrinsic factors. The number of motor units active at any time during a muscle contraction will change the strength of the voltage detected, whilst the diameter of muscle fibres will affect action potential propagation velocity. The depth of active muscle fibres relative to the location of the EMG recording electrode can affect signal amplitude. Both are separated by soft tissues, including subcutaneous tissue, which can act as low-pass filters (Farina et al., 2004). Muscle fibre composition and blood flow can also lead to confounding changes in pH and metabolite removal. Therefore, inconsistencies may occur between actual spinal cord output and recorded EMG amplitude (Moritani et al., 2004).

3.6: EMG Instrumentation

3.6.1: Sampling rate

Sampling rate is a critical factor in the conversion of surface EMG recordings from analogue to digital signals. Applying the Nyquist rate, whereby the sampling rate is at least twice the highest frequency component of the EMG signal, will prevent signal distortion or loss of information; commonly referred to as aliasing (Clancy et al., 2002). The highest frequency components of surface EMG signals for most muscles fall within the range of 400-500Hz (Basmajian and De Luca, 1985, Clancy et al., 2002, Merletti and Hermens, 2004, Winter, 1990). Under-sampling, below the Nyquist rate, can significantly alter measures of EMG amplitude and distort the waveform shape of the EMG signal, through loss of information (Ives and Wigglesworth, 2003). Research has shown that peak amplitude of a rectified EMG signal from triceps brachii during isometric and dynamic contractions significantly decreased when the sampling rate was slowed from 6000Hz to 500Hz and 250Hz (Ives and Wigglesworth, 2003). Slowing the sampling rate from 6000Hz to 1000Hz showed only a minor, non-significant decrease in peak amplitude (Ives and Wigglesworth, 2003), suggesting that oversampling above the Nyquist rate is unnecessary.

In the current thesis, EMG data was sampled at a rate of 1000Hz, meaning 1000 samples were collected per second. Using a sampling rate of 1000Hz conforms to the Nyquist sampling theory and will accurately capture measures of surface EMG amplitude.

3.6.2: EMG electrodes

Surface or indwelling EMG recording electrodes are used to study muscle activity. In the current thesis, surface EMG techniques were chosen to study lower limb muscle activity during quiet standing (Soderberg and Knutson, 2000), in preference to invasive methods. Justification for this technique is that surface EMG provides a 'global' measure of muscle activity, detecting the collective sum of activity in multiple motor units. Invasive fine-wire and needle electrodes are used for detailed analysis of motor units, action potentials from single muscle fibres, or small, deep-lying muscles (Soderberg and Knutson, 2000). Furthermore, the cross-sectional area of muscle from which intramuscular EMG recordings are collected is highly selective and limited to the active area of the needle (Turker, 1993). Surface EMG lacks this degree of selectivity, as surface electrodes can detect electrical activity from adjacent muscles (Soderberg and Knutson, 2000), however this can be minimised through accurate electrode placement.

Non-invasive surface EMG electrodes eliminate potential confounding factors associated with invasive methods including displacement of a needle or wire during muscle contractions, which can cause pain or muscular damage (Soderberg and Knutson, 2000), leading to muscle inhibition or participant non-compliance.

In the current thesis, surface EMG electrodes had a bipolar configuration, whereby two electrodes are placed along the length of the muscle belly and propagation of the action potential between them is recorded (Turker, 1993). A bipolar arrangement was chosen on the basis that action potentials are detected sequentially at each of the electrode sites. The difference in voltage between the two signals represents the amplitude of muscle activity (Kamen and Caldwell, 1996). Any signal common to both electrode sites is attenuated (Kamen, 2004). This procedure is more effective in eliminating surrounding electrical noise, generating a more accurate EMG signal (Richards et al., 2008). Monopolar electrode arrangements can detect surrounding electrical noise and have lower specificity, particularly when recording from small cross-sectional areas of muscle (Turker, 1993).

In the current thesis, EMG electrodes were mounted onto a lightweight casing, with a fixed inter-electrode distance of 20mm. This prevents adjustment of inter-electrode distance relative to the size of the muscle, yet will remain standardised for all muscles being investigated.

3.6.3: Electrode placement

Placement of EMG recording electrodes can be time consuming, requiring preparation of the skin and electrode, accurate measurement from anatomical landmarks using a tape measure, followed by tests of manual resistance to confirm absence of cross talk from adjacent muscles. In the current thesis, all participants will be required to be standing upright during electrode placement procedures. This means the lower limb muscles will be the same length as they would be during quiet standing tests, allowing accurate electrode positioning.

Collecting reliable, repeated measures of surface EMG activity requires standardisation of the position and orientation of recording electrodes over superficial muscles. The European project “Surface EMG for Non-Invasive Assessment of Muscles” (SENIAM) developed standards for electrode placement. These guidelines emerged following a major review of the literature, which aimed to improve standardisation and reproducibility of surface EMG procedures. Within these guidelines, electrode placements are described as a specific point on a line, drawn between two anatomical landmarks. SENIAM recommendations recognise that the optimum longitudinal position of electrodes is between the most distal motor end plate zone and distal tendon (Hermens et al., 2000). Previous research has shown that EMG bipolar electrodes placed over, or within close proximity to, muscle innervation zones, where motor end plates show clustering, nerves terminate and muscle fibres connect, can lower the amplitude of EMG signals (Beck et al., 2008, Rainoldi et al., 2004, Rainoldi et al., 2000). This occurs because motor unit action potentials propagate in both directions, away from the neuromuscular junction. Therefore, when using a bipolar electrode arrangement, signals recorded at each electrode site may succumb to algebraic subtraction, due to the differential amplifier.

This can lead to cancellation of the EMG signal at both electrode sites (Kamen, 2004).

Transversely, electrodes should be located at their maximal distance from adjacent muscles and subdivisions, to reduce the potential for cross-talk (Hermens et al., 2000). Electrode placement recommendations from SENIAM create a reference line between two anatomical landmarks. However, surface EMG recording electrodes also need to be oriented parallel to the direction of the muscle fibres as they are recording propagation of motor unit action potentials along the length of muscle fibres. Therefore, some electrodes require manual rotation prior to attachment to the skin.

The current thesis did not aim to investigate the location of each muscle's innervation zone prior to electrode placement. Therefore, electrode placement for rectus femoris, vastus lateralis, biceps femoris and soleus followed SENIAM guidelines. Electrode placement for vastus medialis, tibialis anterior, peroneus longus and medial gastrocnemius followed alternative procedures implemented in previous literature. All EMG electrode placements will be defined and justified in chapter 4 of this thesis.

3.7: EMG Amplification and Recording

EMG signals are of low power and require amplification before processing. This ensures myoelectric information is not degraded by other electrical, biological or thermal interference, which can affect the true power of an EMG signal (Loeb and Gans, 1986).

Amplification of surface EMG signals should be at least 10 times as high as the electrode-skin resistance (Turker, 1993). Therefore, the amplifier should be set at 1 MΩ or greater (Turker, 1993). In the current thesis, active EMG electrodes contained x350 built-in amplification.

Electrodes mounted in lightweight casing amplifies the signal close to the region where it was detected (Soderberg and Knutson, 2000). Active surface electrodes are not sensitive to changes in electrical resistance between the skin and electrode. This can reduce movement artefacts and optimise the signal to noise ratio (Soderberg and Knutson, 2000). In the current thesis, lower limb EMG activity was recorded using active surface EMG electrodes with 3dB 12-500Hz bandpass. This “roll-off” filter amplifies data within a band of frequencies between the upper and lower 3 dB points in the signal, rather than applying a distinct cut-off threshold (Kamen and Caldwell, 1996). The amplifier should have a frequency bandpass which is equal to, or exceeds, the signal band width (Kamen and Caldwell, 1996, Winter, 1991). The band width or frequency spectrum for surface EMG data is reported to fall within the range of 20-500Hz (Basmajian and De Luca, 1985). However, amplifying the EMG signal may not remove all movement artefacts, thus high- or low-pass filtering may be required.

3.8: Processing the EMG Signal

The raw EMG pattern, detected by surface electrodes, is recorded and visually presented by dedicated software. This signal can be processed using various methods. Raw waveforms can be visually inspected, although this is a highly subjective method lending itself to erroneous recordings in the magnitude of muscle activity (Soderberg and Cook, 1984).

For surface EMG data, low frequency artefacts are suggested to occur between 0-15Hz, with high frequency cut-offs commonly set to 500Hz (Winter, 1991). Merletti and Hermens (2004) suggest surface EMG signals can show movement artefacts or displacement between the electrode-skin interface, within the frequency range of 0-20Hz. Therefore, as will be described in chapter 4 of the current thesis, a high pass filter was applied to the raw EMG signal to attenuate undesirable low frequencies (Turker, 1993).

The raw EMG signal oscillates between positive and negative polarities about the zero line. Therefore, mean values calculated from raw EMG signals would equal zero (Richards et al., 2008). In the current thesis, raw EMG signals were translated to positive polarity using the root mean square function.

Root mean square processing creates a linear envelope, which permits easier identification of changes in the amplitude of muscle activity (Soderberg and Cook, 1984). The root mean square is a one-step process and does not require full-wave rectification. This is because it incorporates the squared values of the raw EMG signal. Each data point is squared then the squares are summed. The sum is then divided by the number of observations, before calculating the square root. This method is suggested to have a sounder mathematical underpinning than processing a low-pass filtered linear envelope (Soderberg and Cook, 1984).

EMG activity is frequently quantified by measuring average EMG activity, which represents the time integral of a root mean squared signal divided by the integration period (Winter, 1991). Therefore in the current thesis, raw EMG signals were processed using the root mean square function, and the average rectified value (Merletti, 1999) calculated by dividing the EMG integral by the time interval.

3.9: Crosstalk

Cross-talk is the unwanted detection of volume conducted signals from muscles other than the muscle of interest (De Luca and Merletti, 1988). This occurs as muscles lie deep, superficial or adjacent to each other, throughout the body. When muscles, lying in close proximity, are activated simultaneously, surface EMG electrodes may detect activity in neighbouring muscles, as they cannot discriminate between the origins of the signals (Kamen and Caldwell, 1996). Therefore, cross-talk can compromise the validity of surface EMG measures. Intra-muscular EMG recording techniques can minimise the risk of cross-talk (Kamen and Caldwell, 1996), yet pose different limitations including pain and muscle inhibition associated with invasive procedures.

Minimising cross-talk in surface EMG measures can be achieved through careful attention to electrode placement and spacing. In the current thesis, accuracy of electrode placement and absence of cross-talk was determined by conducting manual muscle testing against resistance, whilst simultaneously observing raw EMG waveforms (Hermens et al., 2000, Murley and Bird, 2006, Zipp, 1982).

3.10: Physiological Factors Affecting the EMG Signal

3.10.1: Muscular atrophy in older adults

Muscles exhibit structural and functional changes with increasing age. Reduced cross-sectional area and volume of muscle can decrease muscular strength (Porter et al., 1995), whilst loss of motor units can reduce motor unit discharge rate (Erim et al., 1999, Merletti et al., 2002) and alter maximal torque generation (Vandervoort and McComas, 1986). A reduction in the number and size of type II fast twitch muscle fibres can alter the amplitude (Enoka et al., 2003) and rate of muscle force production (Stanley and Taylor, 1993). In the current thesis, smaller muscle mass resulting from age-related muscular atrophy may influence the accuracy of electrode placement and increase the potential for cross-talk in older participants.

3.10.2: Subcutaneous tissue

Excess subcutaneous fat tissue can significantly increase cross-talk in surface EMG recordings (Kuicken et al., 2003), causing attenuation of the signal (Nordander et al., 2003). The thicker the layer of adipose tissue separating the EMG recording electrodes from the muscle fibres of interest, the lower the amplitude of the EMG signal (Kuicken et al., 2003, Nordander et al., 2003). Although skinfold thickness is highly individual, subcutaneous fat and connective tissue replace contractile tissue with increasing age (Porter et al., 1995). This has implications for the accuracy of electrode placement in older participants. Higher amounts of adipose tissue can also increase the risk of movement between the electrodes on the skin surface relative to underlying muscle, increasing the potential for mechanical artefacts and poor reliability of data (Turker, 1993).

3.11: Screening Assessments

3.11.1: Peripheral neuropathy

In the current thesis, all participants were required to have full sensation on the plantar surface of their feet to ensure they perceived the indenting pattern of the textured surfaces. Foot sensation can be assessed using monofilaments (Armstrong et al., 1998), vibration perception thresholds (Perkins et al., 2001), and peripheral neuropathy verbal questionnaires (Armstrong et al., 1998). The latter two methods can be costly, time consuming and may be of limited use in healthy participants. Monofilaments have shown greater sensitivity in detecting foot ulceration compared to vibration perception threshold (Kumar et al., 1991).

The radius, length and elasticity of the nylon monofilament can affect the reliability of the instrument to consistently apply a 10g force to the skin over repeated applications. Research has shown that the buckling force of 10g Bailey Instrument Ltd monofilaments can vary by ±0.5g, with 28% of 50 monofilaments tested, deviating up to ±1.0g (Booth and Young, 2000). However, these tests were conducted using load cells. It is likely that variation in buckling forces may differ when considering the elasticity of the skin.

There is lack of consensus regarding the number and location of sites tested on the plantar and dorsal aspects of the feet, which are sufficient to detect peripheral neuropathy. Intra-tester reliability has shown to be ‘good’ with a 2-site application procedure, ‘moderate’ using 6 sites and ‘poor’ with 1 site (Mawdsley et al., 2004). High specificity has been reported for a 1-site application under the first metatarsal head (Saltzman et al., 2004), with high sensitivity for the third and fourth metatarsal heads (Lee et al., 2003).

Therefore, in the current thesis, screening tests for peripheral neuropathy were conducted using a 10g monofilament (Bailey Instruments Ltd.), as this buckling force is representative of protective foot sensation (McGill et al., 1998). The 10g monofilament was applied bilaterally at 5 sites across the plantar and dorsal aspects of participants’ feet. Inability to detect the monofilament at ≤4 sites per foot (Armstrong et al., 1998, Modawal et al., 2006) signified diminished protective foot sensation, not to be interpreted as complete anaesthesia. This method has been validated in young adults with diabetes (Armstrong et al., 1998), but cannot be generalised to older adults who may present with age-related changes in the nerves, skin perfusion, circulation and sensation. However, monofilaments provide a quick, simple tool to differentiate between sensate and insensate regions on participants’ feet.

3.11.2: Foot posture

The foot is the most distal segment of the lower kinetic chain, therefore changes to its mechanical stability, neuromuscular or proprioceptive function could influence postural control strategies. In particular, age-related physiological changes may alter older peoples foot shape, including greater pronation, limitations to joint movement, toe deformities, toe muscle weakness and reduced tactile sensitivity (Scott et al., 2007).

Healthy adults with pronated (low-arched) feet have shown greater mean deviation of postural sway compared to those with supinated (high-arched) feet (Cote et al., 2005). During single-limb-stance, foot supinators have also shown greater range of CoP movement than participants with neutral foot posture (Hertel et al., 2002). A systematic review by Murley et al (2009) concluded that foot posture can influence lower limb muscle activation patterns during gait, in particular foot inverter and evertor muscles. Foot type has also shown to influence plantar pressure distribution in older people during walking (Scott et al., 2007).

Foot posture can be determined through visual inspection, single or combined anthropometric values, footprint parameters or radiographic evaluation (Razeghi and Batt, 2002). Commonly used clinical assessment tools include Root Theory and Arch Index, which both evaluate foot position based on uni-planar measures of the subtalar joint and calcaneum, and medial longitudinal arch height, respectively (Razeghi and Batt, 2002). The reliability and validity of these measurements have been questioned (McCrory et al., 1997, Menz, 1995, Payne and Richardson, 2000, Saltzman et al., 1995), and in reality tri-planar movement occurs at the foot. Navicular drop assesses pronation based on sagittal movement of the navicular bone using a highly subjective methodology: that being 50% weight-bearing (Razeghi and Batt, 2002). The Valgus Index is complex, time-consuming and has shown poor reliability (Weiner-Ogilvie and Rome, 1998).

Therefore in the current thesis, foot posture was assessed using the Foot Posture Index Version 6 (FPI6) (Redmond et al., 2006). This measurement tool comprises 6-items, classifying foot posture through observation and palpation of the fore-, mid- and rearfoot regions during quiet bipedal standing. Justification for using the FPI6 is that it focuses on previously reported clinical features which are capable of signifying multi-planar changes in foot position (Redmond et al., 2006). Validation studies indicated the FPI6 had better validity and reliability than the original 8-item version, following the removal of two items, those being Helbing's Sign and lateral border congruence (Evans et al., 2003, Keenan et al., 2007, Redmond et al., 2006, Scharfbillig et al., 2004). FPI6 items and scoring system will be described in more detail in chapter 4 of this thesis.

3.11.3: Cognitive ability

Previous research suggests a relationship may exist between high-order cognitive ability and motor performance. Cognitive impairment may be associated with postural instability observed in older adults prone to falling, when performing dual-tasks (Hauer et al., 2003). Common clinical assessment tools include the Abbreviated Mental Test, Six-item Screener, Six-item Cognitive Impairment Test, Clock Drawing Test, Mini-Cog and The General Practitioner Assessment of Cognition (Woodford and George, 2007). However, the Mini Mental State Examination (MMSE) is the most frequently cited tool for evaluating cognitive aspects of psychological function in older people (Folstein et al., 1975).

The MMSE is brief and simple to conduct, and has proven re-test reliability over 24 hours and 28 days with single and multiple administers (Folstein et al., 1975). A maximum score of 30 can be achieved through a series of 11 questions focusing on aspects of orientation to time and place, registration, attention, calculation, recall, language, repetition, and complex commands including basic motor skills (Folstein et al., 1975). The traditional MMSE cut-off score of <24, indicating cognitive impairment in older people (Folstein et al., 1975), is commonly cited throughout literature (Melzer et al., 2004, Menant et al., 2009, Tsang and Hui-Chan, 2005). However, recent evidence suggests individuals scoring ≥27 should be considered cognitively intact (Bravo and Herbert, 1997, Chatfield et al., 2007, O'Bryant et al., 2008, Raji et al., 2009). Using a cut-off score of 27 in highly educated older adults produces a more optimal balance between sensitivity (0.89) and specificity (0.91) values, compared to the traditional cut-off of 24 (O'Bryant et al., 2008). 90% of older adults have shown to be correctly classified for the presence of dementia using a cut-off score of 27 (O'Bryant et al., 2008). The sensitivity of the MMSE to correctly identify cognitive impairment in depressed older people substantially increases (from 8.0% to 37.5%) when using a cut-off of 27, rather than 24 (Raji et al., 2009). This is accompanied by only a minimal decrease in specificity (from 99.4% to 91.3%). Age-adjusted mean MMSE scores from non-demented older adults have shown that those aged over 65 years, commonly score above 27 (Chatfield et al., 2007).

Cognitive test scoring systems do not account for age and intellect, and a number of items within the MMSE are criticised as being too easy (Lopez et al., 2005). Thus, Lopez et al. (2005) recommend the MMSE cut-off score to be set and interpreted by the researcher, in accordance with the study sample. Therefore, in the current thesis, the MMSE cut-off score was set at 27 (participants scoring ≤ 26 were excluded) as older adults were healthy, active, community-dwelling individuals, some with professional occupational backgrounds. MMSE scores were intended to optimise consistency in the cognitive abilities of participants and provide a simple measure of normal versus impaired cognition, rather than the severity of impairment.

3.12: Textured Surfaces

Stimulation of cutaneous receptors, known as mechanoreceptors, may have a significant role in maintaining upright balance, providing information to the CNS about the position of the body relative to the ground (Kennedy and Inglis, 2002). When the surface of the skin is indented by a rough texture or object, such tactile stimulation may change the pattern of afferent activity in populations of mechanoreceptors (Johnson, 2001). As the pressure increases, a larger area of the skin is indented, irrespective of the size of the stimulus. However, the downward vertical pressure applied to all conditions will be the same, equating to the bodyweight of the participant and constant force of gravity on top of the surface. It has been suggested that plantar mechanoreceptors may be sensitive to changes in the distribution of pressure under the feet during standing (Kavounoudias et al., 1998). Therefore, a textured surface providing sensory input to plantar mechanoreceptors may be capable of altering peripheral neural activity, and thereafter sensorimotor control of balance.

It is possible that the number of protrusions indenting the plantar surface of the foot (per square unit), or the prominence of the indenting protrusions specific to a textured pattern, may deliver differing degrees of tactile stimulation, thereafter determining the magnitude or direction of change observed in kinetic or kinematic variables. As discussed in chapter 2 of this thesis, previous research has shown that textured surfaces and foot insoles can significantly alter balance performance and muscle activation independently, compared to control. Irrespective of differences in methodologies between previous studies, these changes in balance or EMG activity may be unique to the geometric pattern of each textured intervention. Different textures may bring about different effects on measures of postural sway and lower limb muscle activity: such postulations have not yet been explored. Therefore, in the current thesis, two different textured surfaces, and one surface as control, were investigated on the basis that their variety of indentation, and thus the degree to which each textured pattern depressed the skin on the plantar surfaces of the feet, may bring about differing effects on postural sway and lower limb EMG activity.

3.13: Methods of Analysis

3.13.1: Postural sway and lower limb EMG analysis

Previous research indicates that measures of sway amplitude and frequency are not consistent over the course of a 60 second quiet standing test in adults aged 20-60 years (Samson and Crowe, 1996). During quiet standing with feet together and eyes open, mean sway values in the first 15 seconds were greater than subsequent 15 second periods. This was also true for quiet standing with eyes closed (Samson and Crowe, 1996).

The findings of Le Clair and Riach (1996) also suggest the first 10 seconds of quiet standing may capture highly variable sway data representing re-stabilisation of participants. Raymakers et al. (2005) attribute such variability to stabilisation of the force platform. For these reasons, some previous studies have discarded the initial period of sway data from analysis (Wilson et al., 2008). Mochizuki et al. (2006) investigated CoP movement in healthy young adults, when standing quietly on a force platform and a variety of supporting surfaces of different sizes, over 32 seconds. Only the first and last 2 seconds of sway data were discarded to eliminate errors associated with filtering procedures (Mochizuki et al., 2006).

In the current thesis, the first 10 seconds of sway data were not discarded from analysis. This can be justified as previous research suggests short-term variability of CoP movement may represent exploratory activity of the CNS (Mochizuki et al., 2006, Riley and Turvey, 2002). Once the boundaries of the base of support have been located, only then may the CoP move within a more refined range (Mochizuki et al., 2006). It is possible that the first 10 seconds of quiet standing may provide rich data relating to the effect of texture on exploratory CoP movement (Mochizuki et al., 2006, Riley and Turvey, 2002). Furthermore, it is more likely that adults could experience loss of balance during, rather than following, the re-stabilisation process after standing up from a seated position.

Chapter 4 of the current thesis will describe the data processing and extraction procedures, whereby postural sway and lower limb EMG data are reported over three different time intervals including: the first 10, latter 20 and overall 30 seconds of quiet standing.

3.14: Summary

In order to test the hypothesis and implement evidence-based practice, the following methods will be used, based on appraisal of current literature.

Quiet standing balance is a sufficiently robust balance task to assess the effects of different textured surfaces on postural stability in healthy adults. Sway parameters, including AP and ML SD, AP and ML range and CoP velocity will be extracted from a Kistler force platform. Sway data will be collected over three different time intervals: the first 10, latter 20 and overall 30 seconds of quiet standing. This will enable exploration of the effect of enhanced tactile stimulation on balance performance during and after participant re-stabilisation.

Measures of EMG amplitude will be collected using surface EMG techniques, including bipolar recording electrodes. EMG electrode placement and orientation will be standardised in line with previous recommendations. Consideration will be given to levels of adipose tissue and signs of muscular atrophy which could affect the accuracy of electrode placement. EMG processing techniques will involve rectification and high-pass filtering of the raw signal, before calculating the amplitude of muscle activity.

Participant screening assessments will include foot posture using the FPI6, foot sensation using monofilaments, and for older participants, the MMSE to determine cognitive ability.

Two different textured surfaces, and a smooth surface as control, will be investigated as the geometric pattern specific to each surface may bring about different effects on plantar mechanoreceptor activity.

CHAPTER 4: METHODOLOGY FOR THE EFFECT OF TEXTURED SURFACES ON POSTURAL SWAY AND LOWER LIMB MUSCLE ACTIVITY DURING QUIET STANDING IN HEALTHY YOUNG ADULTS

4.1: Introduction

This chapter will describe the methods and procedures implemented during the investigation of the effect of textured surfaces on standing balance and lower limb muscle activity in healthy young adults, in the current thesis. Key issues will include participant recruitment, instrumentation (including the Kistler force platform and EMG) and study procedures including participant screening assessments, standardisation of body position and apparel during testing.

4.2: Background

Health care professionals frequently prescribe footwear interventions as a treatment modality for the prevention and rehabilitation of overuse musculoskeletal injury in young adults, including Achilles tendonitis (Wallace et al., 2004), iliotibial band syndrome (Gross and Napoli, 1993), patellofemoral pain syndrome (Saxena and Haddad, 2003), plantar fasciitis (Landorf et al., 2006), posterior-tibial tendon dysfunction (Alvarez et al., 2006) and medial tibial stress syndrome (Rome et al., 2005). Footwear interventions have been reported to increase comfort and improve proprioception in both athletic and high-risk groups.

The exact biomechanical mechanism of action of footwear interventions remains unknown, yet they are commonly prescribed to improve lower extremity alignment by means of controlling abnormal or excessive movement at the subtalar joint. Some authors consider the shock-absorbing effects of footwear intervention to be their most useful asset (Ball and Afheldt, 2002, Nigg et al., 1999). However, a growing body of literature suggests footwear interventions may alter plantar afferentation patterns which thereafter influence sensorimotor function during daily and sporting activities (Corbin et al., 2007, Hertel et al., 2005, Mundermann et al., 2006, Nurse et al., 2005, Palluel et al., 2008).

In particular, textured floor surfaces and foot insoles have been shown to alter balance performance (Corbin et al., 2007, Palluel et al., 2008, Watanabe and Okubo, 1981), lower limb muscle activity (Nurse et al., 2005), and discrimination of ankle joint position (Waddington and Adams, 2000, Waddington and Adams, 2003) in young adults.

To date, no study has investigated the effect of different textured surfaces on measures of postural sway and lower limb muscle activation during quiet standing in healthy young adults. This study will generate vital evidence concerning the future potential for textured surfaces or foot insoles to be used as interventions for improving balance performance in healthy, athletic, injured or pathological groups of young adults.

4.3: Ethical Approval

Ethical approval was granted by the School of Health and Social Care Research Governance and Ethics Committee at Teesside University. Verbal and written informed consent was obtained from all participants meeting the inclusion criteria (Appendix 2).

4.4: Recruitment and Sampling Method

Recruitment posters (Appendix 3) were located within the physiotherapy laboratories at Teesside University, requesting that any potential participants voluntarily contact the researcher (ALH) for further information (Appendix 4). The researcher also attended the physiotherapy laboratories on various occasions to describe and explain the study to potential participants. Any young adults who fit the inclusion criteria and were interested in participating were asked to speak with the researcher.

4.5: Participants, Inclusion and Exclusion Criteria

A convenience sample of 24 healthy young adults (7M, 17F) from Teesside University and the local community was recruited. Any participant presenting neuromuscular disease, peripheral sensory neuropathy, type II diabetes, an inner ear disorder, vertigo, dizziness, inability to walk unassisted for 10 metres or inability to stand up and sit down without using their hands, were excluded from the study. Basic demographic data including weight, height, body mass index (BMI), eye height and knee height were collected for all participants. Measuring each participant's eye height enabled the visual target to be mounted on a board at eye level, and thus standardise head position to prevent vestibular disruption. Knee height was measured for each participant in order to standardise the height of the plinth and therefore lower limb joint angles before standing up from a seated position. Screening assessments for healthy young adults included a short written questionnaire detailing current health and fitness, including any surgery undergone in the previous 12 months, assessment of foot posture using the FPI6 (Redmond et al., 2006) and bilateral foot sensation using monofilaments (Appendix 5).

4.6: Research Environment

All tests were conducted in the physiology laboratory at Teesside University. The temperature of the laboratory remained constant at approximately 22ºC throughout all testing procedures, to prevent changes in lower limb EMG activity and force associated with variations in ambient room temperature (Bell, 1993).

4.7: Instrumentation

4.7.1: Kistler force platform

Data for postural sway during quiet standing was obtained from a Kistler force platform (Model 9286AA, Kistler, Alton, UK), and sampled at a rate of 1000Hz (Bergland and Wyller, 2004).

4.7.2: Electromyography

All surface electromyography (EMG) recordings were collected using a 16-channel Biopac system (Model MP100), using active surface EMG recording electrodes (Type TSD150B, 11.4mm diameter, electrode spacing 20mm) (Fig 4.1), with 3dB 12-500Hz bandpass and x350 built in amplification. EMG data was processed using dedicated AcqKnowledge software (Version 3.7.3, BIOPAC Systems, Inc.), which was compatible with the EMG instrumentation. All EMG recordings were sampled at a rate of 1000Hz (Merletti and Hermens, 2004).

Fig 4.1: EMG recording electrode

4.7.3: Skin preparation

Participant's skin on the dominant lower limb was shaved using an electric razor, only if hair was overlying the skin surface, at which the electrodes were to be placed. The skin was cleaned with isopropyl alcohol at eight electrode sites and allowed to vaporise prior to placing the electrodes. Participants were informed this procedure would be pain-free and not cause any discomfort. Hypoallergenic conducting gel was applied to the electrodes to ensure optimal electrical contact with the skin surface. Bipolar surface EMG recording electrodes were placed over superficial muscles on the dominant lower limb. The dominant leg was determined by asking participants 'with which leg would you kick a football?' EMG recording electrodes were then attached to that leg. A pre-gelled ground reference electrode (Blue Sensor ®) was placed at the tibial tuberosity on the non-dominant lower limb. As the tibial tuberosity is a non-muscular site and located close to the recording electrodes when the lower limbs are parallel during bipedal stance, optimal grounding would therefore be achieved (Turker, 1993).

4.7.4: Electrode placement and orientation

In healthy young adults, the eight muscles of interest were vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA), peroneus longus (PL), medial gastrocnemius (MG) and soleus (SOL) (Fig 4.2 and 4.3).

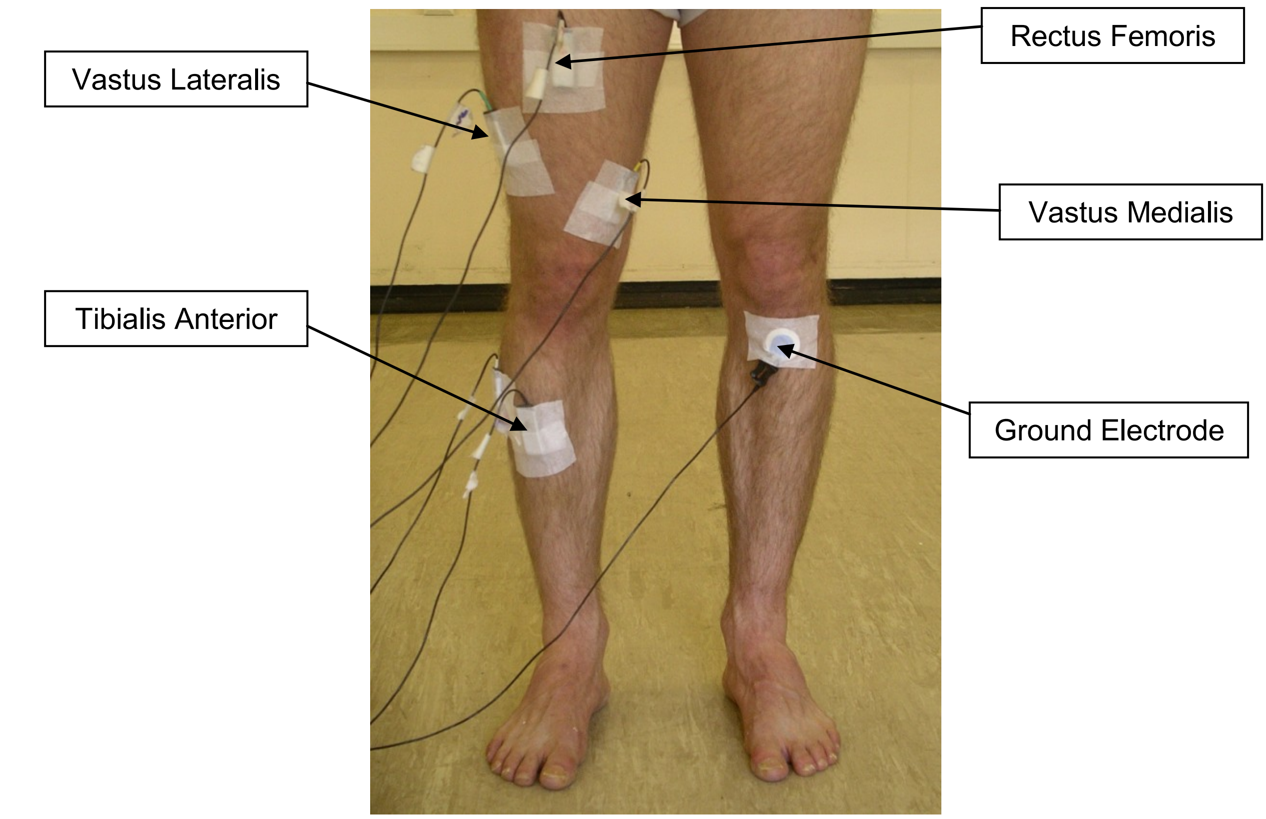

Fig 4.2: EMG electrode placement for healthy young adults – anterior view

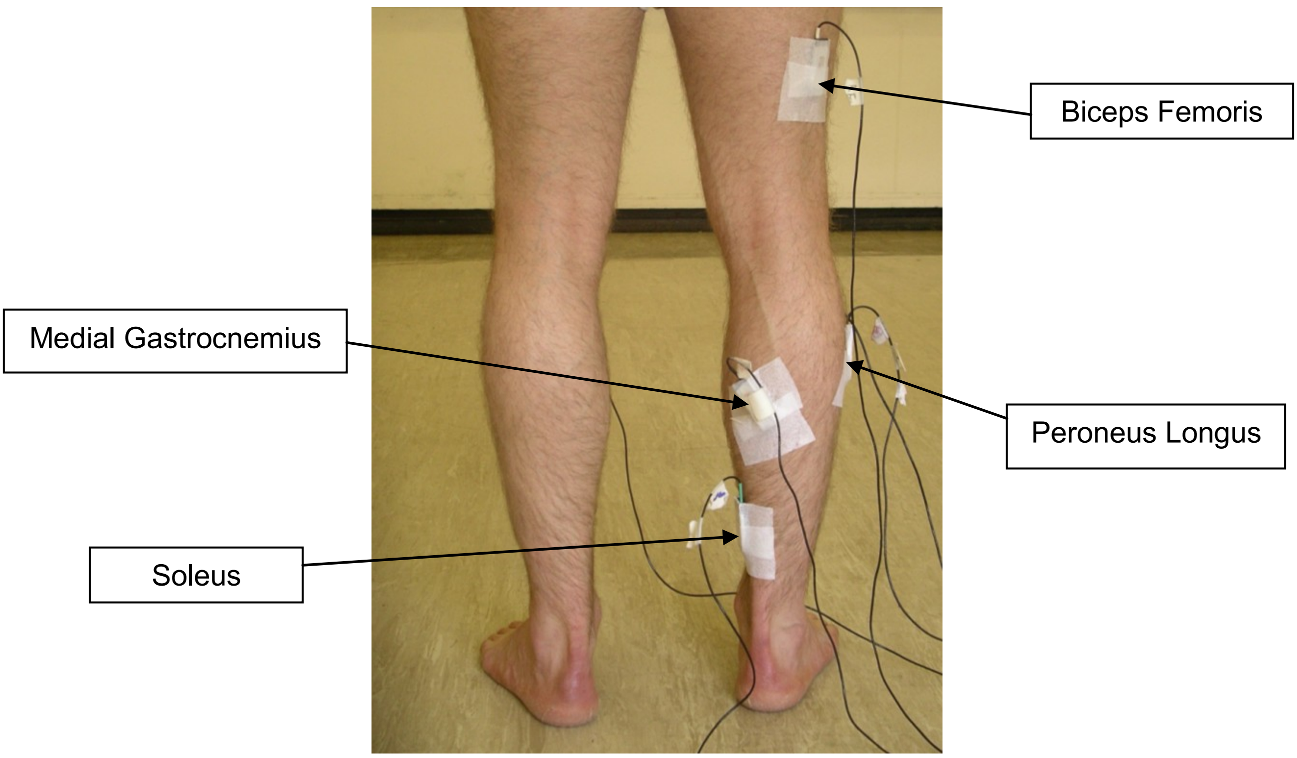

Fig 4.3: EMG electrode placement for healthy young adults – posterior view

Electrode placement and orientation for all muscles were standardised, following recommendations from SENIAM or procedures reported in previous literature.

Vastus Medialis: VM comprises distal (vastus medialis oblique) and proximal (vastus medialis longus) portions, which can be differentiated by their fibre alignment (Lieb and Perry, 1968, Ono et al., 2005). The distal fibres are reported to be oriented 50-55º medially, and the upper fibres 15-18º medially, in the frontal plane (Lieb and Perry, 1968). In the current study, the VM electrode was placed at 3-4cm above the superior-medial border of the patella (Cram and Kasman, 1998, Gilleard et al., 1998) and oriented at a 55º angle to the long axis of the femur (Lieb and Perry, 1968) in order to capture EMG activity in vastus medialis oblique, given its exclusive role in providing medial stabilisation of the patella during knee extension (Hubbard et al., 1997). However, it is possible that electrical activity recorded from VM may represent both portions working in synergy (Zipp, 1982).

Rectus Femoris: the electrode was positioned at 50% on the line from the anterior superior iliac spine to the superior part of the patella and oriented along this same line (SENIAM).

Vastus Lateralis: the electrode was oriented at 15º laterally to the longitudinal axis of the femur and placed at two thirds on the line from the anterior superior iliac spine to the lateral border of the patella (SENIAM).

Biceps Femoris: the electrode was placed at 50% of the distance between the ischial tuberosity and head of the fibula (LeVeau and Andersson, 1992, Nawoczenski and Ludewig, 1999, SENIAM).

Tibialis Anterior: the electrode was placed 25% of the distance between the lateral knee joint space and the lateral malleolus, just lateral to the tibia (Leis and Trapani, 2000, LeVeau and Andersson, 1992, Nawoczenski and Ludewig, 1999, VonTscharner et al., 2003, Zipp, 1982). Given the close proximity of TA to PL, this alternate procedure for TA electrode placement was followed, rather than SENIAM guidelines, on the basis that it may provide a more reproducible, accurate reference line.

Peroneus Longus: the electrode was placed 5-7cm distal to the lateral head of the fibula along the lateral aspect of the fibula (Leis and Trapani, 2000), oriented in the direction of the line between the head of the fibula and lateral malleolus (SENIAM). Care was taken to place the PL electrode parallel to the fibula (Leis and Trapani, 2000), due to the risk of cross talk between TA and PL. Justification for following procedures implemented by Leis and Trapani (2000), rather than SENIAM recommendations, is that precise measurements from an anatomical landmark may enhance the reliability of electrode placement, compared to those based on leg length percentage.

Medial Gastrocnemius: the electrode was placed over the most prominent bulge of the medial muscle belly (SENIAM) and medially oriented at an angle of 15º to be aligned with muscle fibre pennation (Narici et al., 1996) when the ankle is positioned at right angles to the foot. A reference line from the medial aspect of the Achilles tendon insertion to the medial side of the popliteal cavity (Rainoldi et al., 2004) was used to guide electrode orientation. The MG electrode was manually rotated to match muscle fibre pennation during quiet bilateral standing.

Soleus: the electrode was placed over the medial inferior fibres, in a region where gastrocnemius was not superficial to SOL (Laughton et al., 2003), just distal to MG and medial to the Achilles tendon (Cram and Kasman, 1998, SENIAM). The SOL electrode was oriented in the direction of the line between the medial condyle to the medial malleolus (SENIAM). Although SOL has limited surface area where it lies superficially, in the current study, the SOL electrode was placed on the distal half of the muscle, guided by palpation.

4.7.5: Manual muscle testing

To confirm the accuracy of electrode placement, participants were seated on the edge of a height-adjustable plinth and asked to perform the following movements, against manual resistance:

VM – Resisted knee extension
RF – Resisted knee extension
VL – Resisted knee extension
BF – Resisted knee flexion
TA – Resisted ankle dorsiflexion with inversion
PL – Resisted ankle plantarflexion with eversion
MG – Resisted ankle plantarflexion (rising onto the toes)
SOL – Participants were asked to lift the heel from the floor, against manual resistance pushing downwards on the knee.

4.8: Triggering

Postural sway and EMG data were synchronously recorded over a 30 second window. A trigger connection between the Kistler force platform and Biopac EMG instrumentation ensured there was no time delay in the commencement of data acquisition for each system.

4.9: Participant Positioning

4.9.1: Foot positioning and foot templates

Foot positioning was standardised for all participants during each test of quiet standing using individual paper foot templates (Buckley et al., 2005). A sheet of plain white paper was placed over the top surface of the force platform. The participant was then asked to stand barefoot, on the centre of the force platform on top of the paper in their normal, preferred angle-base of support (Panzer et al., 1995, Stel et al., 2003). To prevent influencing foot position, it was not stipulated whether or not the feet should be placed parallel. The researcher drew around and thereafter manually cut out the outline of participants' barefeet to create an individual foot template (Buckley et al., 2005, Du Pasquier et al., 2003). With the participant seated on the end of the plinth, this template was laid over the textured and control surfaces prior to the commencement of each trial then removed once the feet had been correctly positioned. Participants were asked not to displace their feet from this position whilst rising from a seated position, when standing fully upright or upon sitting down. The template was re-laid on the textured and control surfaces following each trial to reposition the feet (Figure 4.4).

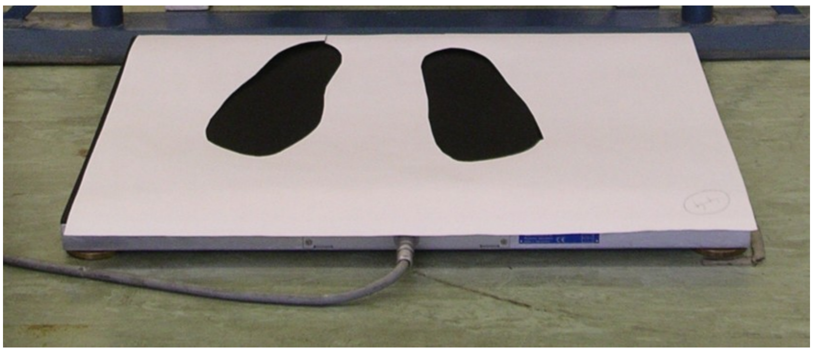

Fig 4.4: Foot template

4.9.2: Upper limbs

To minimise individual variability of upper body movement during quiet standing, all participants were instructed to stand with their upper body relaxed and arms hanging by their sides (Laughton et al., 2003, Mochizuki et al., 2006).

4.9.3: Head orientation

When standing quietly with eyes open, participants were instructed to look straight forwards and focus on the middle of a visual target, to prevent vestibular disruption (Brandt et al., 1981). A black circular visual target of 100mm diameter was mounted onto a board positioned 3m from the centre of the force platform (Fig 4.5). The visual target was adjusted to the eye level of each participant and was therefore placed at a height equal to eye level, plus the 350mm depth of the force platform and textured or control surface combined (Rome and Brown, 2004).

Fig 4.5: Visual target mounted at participants' eye height

4.9.4: Seating specifications

Between tests of quiet standing balance participants sat on the end of a hydraulic plinth, which provided a height-adjustable seat without trunk/back support or armrests. Plinth height was standardised to knee height, defined as 100% the length of the dominant lower leg, from the centre of the knee joint to the ground when standing barefoot (Khemlani et al., 1999). The height of the sitting surface was measured using a tape measure from the ground to the upper surface of the plinth and adjusted to each participant's knee height plus an additional 350mm to account for the depth of the force platform and textured or control surface combined.

4.10: Screening Assessments

4.10.1: Peripheral neuropathy

Sensory screening tests for peripheral neuropathy were undertaken for all participants prior to recruitment. A 5.07 retractable nylon monofilament (Retractable Monofilament with Duraban®, Bailey Instruments Ltd., WO21864 2010-07) with a buckling force of 10g was applied bilaterally at 5 sites on the plantar and dorsal surfaces of the feet. Applying the monofilament perpendicular to the surface of the skin with a steady pressure will cause it to bend and exert a specific force of 10g.
Prior to using the monofilament on participants' feet, the nylon wire was pressed and buckled against the dorsum of their hand three times. This allowed for familiarity with the gentle pressure which was to be identified on the feet. Additionally, this brand of monofilament has shown a decline in buckling force during the first three compressions before reaching a steady state of force (Booth and Young, 2000).

The healthy young adults were asked to lie prone on a plinth with their feet overlying the plinth, and therefore in a position whereby they could not observe the monofilament being applied. All participants were asked to close their eyes during monofilament testing. A monofilament was applied perpendicularly to five testing sites on the left and right foot. Care was taken to apply the monofilaments to non-callused areas on the feet. The monofilament was applied with a steady pressure and held for a period of approximately 1.5 seconds at each site commencing from the point at which the filament buckled (Mawdsley et al., 2004). This time period was determined by the tester rather than using a stop clock. Upon identification of the stimulus, participants were instructed to respond by saying 'yes'. A maximum of three applications were permitted at each testing site. If no response was given following three applications per site, this area was marked as 'no response'. An inability to identify the monofilament stimulus at selected regions of the foot did not indicate the presence of complete anaesthesia or peripheral neuropathy, but rather suggested diminished sensation at a specific site. Due to time constraints it was not possible to assess peripheral sensation using monofilaments of varying force. This could have shown that participants may have various sensory thresholds over the plantar aspect of the foot. Bilateral monofilament testing was conducted over a period of approximately 5-10 minutes. Each site was tested only once per foot. The monofilament was applied to the following sites bilaterally:

1. Dorsal Surface – an area between the first and second toes
2. Plantar Surface – hallux
3. Plantar Surface – first metatarsal head
4. Plantar Surface – fifth metatarsal head
5. Plantar Surface – heel

Throughout all test procedures, two Bailey Instrument Ltd. monofilaments were used interchangeably, on alternate days. Research has shown that after 100 consecutive applications, the buckling force of a 10g Bailey Instrument Ltd. monofilament can deteriorate to <9g (Booth and Young, 2000). Therefore to ensure a 10g buckling force was applied to all participants, each monofilament was given 24 hours recovery period as suggested by Booth and Young (2000).

4.10.2: Foot posture index

Foot posture was assessed for all participants using the FPI6 (Redmond et al., 2006), in order to determine the extent to which their feet were pronated, supinated or in a neutral position. Whilst participants were stood barefoot in their normal, comfortable standing position, with feet approximately shoulder-width apart, the researcher subjectively quantified bilateral foot position using six measures. These measures included observations and palpation of the; position of the talar head, supra and infra lateral malleoli curvature, degree of calcaneal inversion/eversion in the frontal plane, talo-navicular prominence, height of the medial arch and the extent of abduction/adduction of the forefoot on the rearfoot (Redmond et al., 2006). Each item was scored individually for each foot on a scale of -2 to +2. Scores of 0 are considered to represent neutral positioning; a minimum score of -2 indicates supination and a maximum score of +2 implies pronation. In accordance with reference values proposed by Redmond et al. (2006) participant's scores for each foot were categorised as follows: normal (0 to +5); pronated (+6 to +9); highly pronated (10+); supinated (-1 to -4) and highly supinated (-5 to -12). Participants were then classified as showing normal foot positioning, bilateral pronation, unilateral pronation, bilateral supination or unilateral supination.

4.11: Textured Surfaces

Two different textured surfaces, and one smooth surface as control, were used (Algeos UK Ltd., Liverpool, United Kingdom) (Fig 4.6). Texture 1 (T1) (Evalite Pyramid EVA, 3mm thickness, shore value A50, black, OG1549) and Texture 2 (T2) (nora®Lunsasoft Mini Non Slip, 3mm thickness, shore value A50, black, OG2250) were selected from the range of EVA soling materials. T1 had small, pyramidal peaks with centre-to-centre distances of approximately 2.5mm, whilst T2 displayed convex circular patterning with centre-to-centre distances of approximately 5mm. The control texture (C) (Medium Density EVA, 3mm thickness, shore value A50, black, OG1304) was chosen from the range of plain EVA and had a completely flat surface with no indentations. The textured surfaces were cut manually from commercially available sheets to smaller dimensions of 428mm x 630mm x 3mm, in order to cover the top surface of the Kistler force platform.

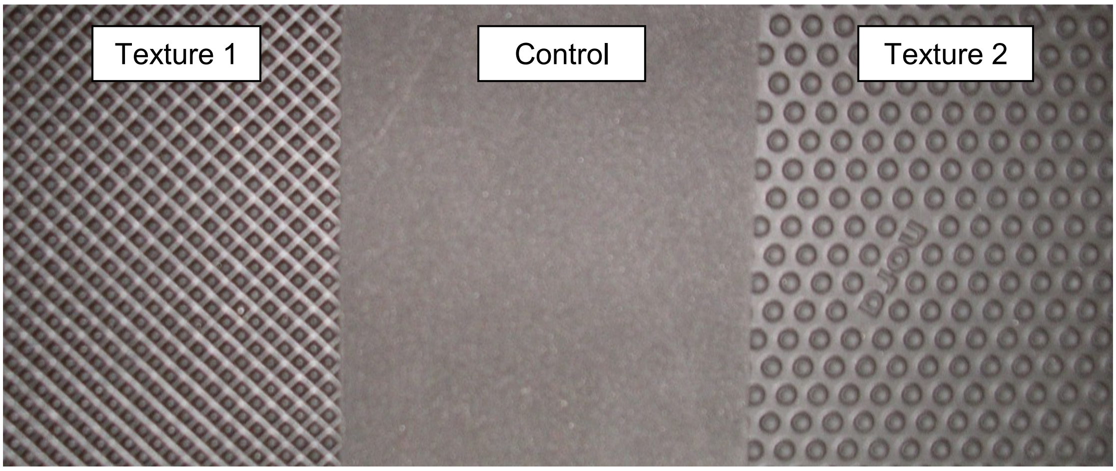

Fig 4.6: Textured and control surfaces

4.12: Procedures

4.12.1: Clothing

All participants were requested to wear loose fitting shorts, a skirt or to roll their trouser leg up to the level of the thigh. Participants were instructed to remove all footwear, socks, tights/stockings and to remain barefoot during data collection.

4.12.2: Randomisation of conditions

The textured and control surfaces and two visual conditions were randomised using number cards selected by participants: 1 = Texture 1, eyes open; 2 = Texture 1, eyes closed; 3 = Texture 2, eyes open; 4 = Texture 2, eyes closed; 5 = Control, eyes open; 6 = Control, eyes closed.

4.12.3: Quiet standing balance

Participants were asked to sit on the end of the plinth which had been adjusted to their knee height. Prior to standing, the foot template was placed over the force platform and textured surface, to allow participants to position and standardise their feet correctly (Du Pasquier et al., 2003). Participants were then asked to stand up, keeping their feet in the position determined by the template (Fig 4.7). During quiet standing balance tests, participants focussed on the middle of a black, circular visual target, which was mounted on a board 3m away from the centre of the force platform and at eye level.

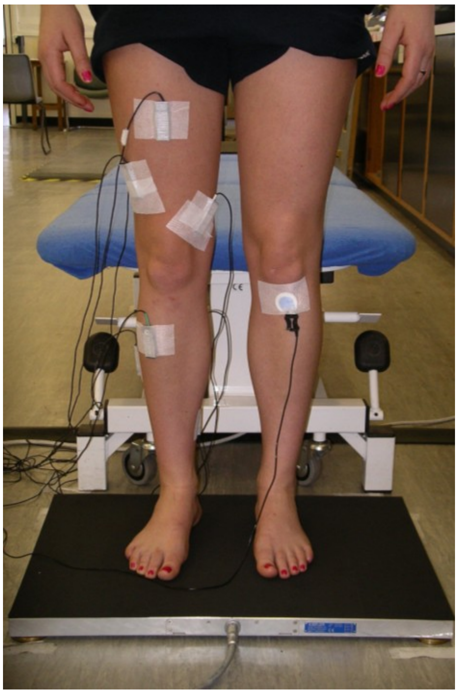

Fig 4.7: Quiet standing balance

Testing procedures began once the participant was correctly positioned – the trunk and lower limbs were perceived to be vertical. Immediately upon reaching this position, the trigger was pressed, which began data collection simultaneously from the Kistler force platform and EMG system.

All participants were tested in barefoot, bipedal quiet standing with eyes open and standing with eyes closed, for an interval of 30 seconds, as indicated by the researcher. This was conducted under all texture and no texture conditions. Healthy young adults conducted three repetitions with eyes open and three repetitions with eyes closed for each surface, therefore 18 repetitions in total.

4.12.4: Rest periods

Following each test of quiet standing balance, participants were asked to sit down and take their feet off the textured surfaces and force platform. All participants sat quietly in their normal comfortable position, with their feet resting on either side of the force platform, on the laboratory floor, for two minutes between all tests. This rest period served to prevent habituation to the sensory stimulus of the textured indentations, whilst allowing the next test condition to be prepared and the force platform re-calibrated, prior to data collection.

4.13: Data Processing and Extraction

Postural sway parameters, including AP and ML SD, AP and ML range were extracted automatically from the Kistler force platform, using the Bioware software package. Mean CoP velocity was calculated according to the formula used by Raymakers et al. (2005) (see below), where Vd is displacement velocity and *n* the number of samples:

CoP velocity (mm s^{-1}): $\frac{\sum Vd}{n}$

Raw EMG data was high-pass filtered at 20Hz and processed using a root mean square moving window of 30ms. For each muscle, the average rectified value was calculated by dividing the EMG integral by the time interval.

The current thesis did not normalise lower limb EMG data as a within-subjects study design was used where participants were exposed to all experimental and control conditions, thus acting as their own controls (Edwards et al., 2008). Normalisation is required when the amplitude of EMG signals are to be compared between-subjects or between-sessions, in order to identify proportional changes in muscle activity relative to a reference value: this was not the case in the current thesis (Soderberg and Knutson, 2000).

In healthy young adults, for both postural sway and EMG parameters, the mean of three repetitions of quiet standing was calculated for all participants. This mean value was used in the analysis. Upon identifying a corrupt data file, which could not be analysed, mean sway and EMG scores were calculated using data from the remaining viable trials.

Mean sway parameters and average lower limb muscle activity was collected over 30 seconds of quiet standing, with eyes open and eyes closed. Postural sway data for the first 10 and latter 20 seconds of quiet standing were also calculated for both visual conditions. Only when textured surfaces showed to have a significant effect on sway parameters, was EMG data reported for these same split time intervals.

4.14: Statistical Analysis

Data were analysed using the Statistical Package for Social Sciences (SPSS, Chicago, IL, USA) version 13.0. For each of the postural sway and EMG variables, a repeated measures analysis of variance (ANOVA) was carried out to determine statistically significant differences among the textured and control conditions with an alpha level set at 0.05. Post-hoc statistical analysis was also conducted using pair-wise comparisons to identify the magnitude of change in variables between conditions. The tables of results will report means and 95% CI of differences between each pair of conditions, *p* values from the pairwise comparisons and percentage change in variables, between control and textured conditions. Percentage change in postural sway and EMG variables between the two different textured surfaces were not presented as this was not considered to be clinically important. The aim of this thesis was to explore how two different textured surfaces altered balance performance and muscle activity relative to ‘no texture’. This overall approach allowed identification of any clinically relevant changes in balance and muscle activity when standing on a textured surface.

In the current thesis tests for normality were not conducted, and all data was analysed using parametric tests. This can be justified as Field (2005) indicates statistical tests for normality cannot determine whether a significant deviation from normal distribution is of sufficient magnitude to bias the application of parametric or non-parametric statistical tests to data. Furthermore, Hopkins et al. (2006) suggest parametric tests are sufficiently robust and sensitive for use with both normally and non-normally distributed data set. The repeated measures ANOVA is considered robust enough to cope with most deviations from normality: it is of greater importance that data meets the assumption of sphericity (Field, 2005). Where the assumption of sphericity was violated, a Greenhouse-Geisser correction was applied.

All postural sway and lower limb EMG data from the repeated measures ANOVAs were reported without a Bonferroni adjustment. The purpose of a Bonferroni adjustment is to decrease the likelihood of Type I errors which can arise when performing multiple tests. However, Bonferroni adjustments are criticised for being too aggressive, increasing the potential for Type II errors and for true differences to be reported as non-significant (Morgan, 2007, Perneger, 1998, Rothman, 1990).

4.15: Within-Session Reliability

The within-session reliability of postural sway and EMG data reported in this thesis was explored using intra-class correlation coefficients (ICCs) and the standard error of measurement (SEM). The reliability of data was assessed for the baseline control condition for three repeated trials of quiet standing over 30 seconds with eyes open only.

Relative reliability for postural sway and EMG variables were calculated using $ICCs_{(3,1)}$ and interpreted according to the following definitions: 0 to 0.1 show negligible reliability, 0.11 to 0.40 are slightly reliable, 0.41 to 0.6 show fair reliability, 0.61 to 0.8 have moderate reliability and above 0.81 indicates substantial reliability (Shrout, 1998, Shrout and Fleiss, 1979). Thus an ICC of 1 would indicate perfect reliability.

Absolute reliability was determined by calculating the SEM, also known as the typical error (Batterham and George, 2003). The SEM can be calculated using two different methods. The first is to conduct calculations using the standard deviation of scores from a large study sample, (obtained from the ANOVA), together with the reliability coefficient, that being the ICC score: $SEM = SD \times \sqrt{(1 - ICC)}$. However, different levels of ICCs exist, and therefore the SEM value will be dependent upon the ICC chosen, which could lead to inconsistencies (Weir, 2005). In the current study, the SEM was determined by calculating the square root of the ANOVA error variance (Hopkins, 2000, Stratford and Goldsmith, 1997): $SEM = \sqrt{residual}$.

CHAPTER 5: RESULTS FOR THE EFFECT OF TEXTURED SURFACES ON POSTURAL SWAY AND LOWER LIMB MUSCLE ACTIVITY DURING QUIET STANDING IN HEALTHY YOUNG ADULTS

5.1: Introduction

This chapter will present the results for the effect of textured surfaces on postural sway parameters and lower limb EMG activity during quiet standing balance with eyes open and eyes closed, in healthy young adults. Measures of postural sway will be reported over 30 seconds, thereafter split into time intervals including the first 10 and latter 20 seconds. Data collected for lower limb EMG activity will be presented during quiet standing over 30 seconds under both visual conditions. Only when textured surfaces had a statistically significant effect on postural sway over the first 10 or latter 20 seconds of quiet standing tests, will the corresponding EMG data be explored.

As justified in chapter 4 of this thesis, the distribution of data was not explored using tests of normality. However, additional descriptive statistics, including mean, SD, median and inter-quartile range, for data presented within this chapter can be seen in Appendices 6-13, providing insight into their distribution.

This chapter will conclude, reporting the within-session reliability of postural sway and lower limb EMG data collected from healthy young adults. Reliability analysis was conducted for tests of quiet standing on the control surface over 30 seconds with eyes open. This exploratory analysis was conducted after the main investigation, to provide greater insight into the variability of data in a healthy young population.

5.2: Healthy Young Adults

A convenience sample of 27 healthy young adults, were recruited from Teesside University and the local community. Three participants from the 27 that were recruited were excluded from the study as seen in the Consolidated Standards of Reporting Trials (CONSORT) flow diagram (Figure 5.1). One participant failed to attend their scheduled testing session, on multiple occasions. Two further participants withdrew from the study prior to data collection, due to other time constraints. Descriptive characteristics of the healthy young adults and the results of the FPI6 and monofilament screening tests can be seen in Table 5.1.

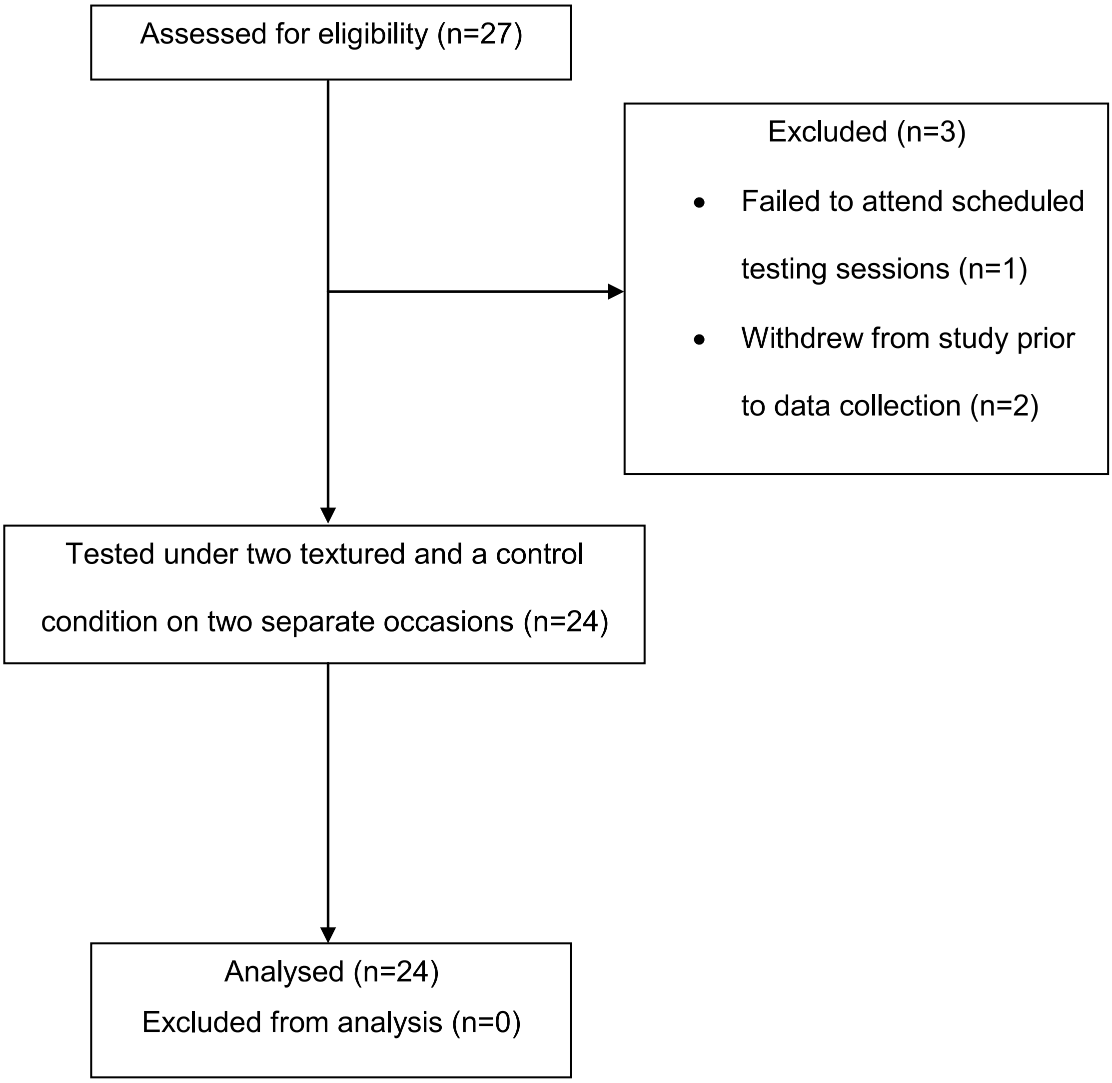

Figure 5.1: CONSORT flow diagram for healthy young adult recruitment and exclusions

Table 5.1: Descriptive characteristics of healthy young adults (n=24)

Gender	Age (yrs) Mean (SD)	Height (cm) Mean (SD)	Weight (kg) Mean (SD)	BMI (kg/m^2) Mean (SD)	FPI6 (n=)	Monofilament Testing Mean (SD) (maximum score: 10)
7 M, 17F	27.7 (7.9)	167.0 (6.0)	67.1 (11.8)	23.8 (4.1)	Normal (22) Bilateral Pronation (1) Bilateral Supination (1)	10.0 (0.0)

5.3: Postural Sway during Quiet Standing in Healthy Young Adults

5.3.1: Postural sway over 30 seconds with eyes open in healthy young adults

Table 5.2 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during quiet standing over 30 seconds with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CIs for C-T1 for AP sway parameters are less conclusive. The CIs for C-T1 for AP SD and AP range both show a marked asymmetry (zero close to the upper limit), in the direction of a decrease in sway. The CIs for T1-T2 for both AP SD and AP range do not contain zero.

Whilst the asymmetries between C-T1 for AP SD and AP range do not reach significance, these observations do tentatively suggest T1 may have the capacity to reduce AP sway. Relative to C, T1 reduced AP SD by 6.8% and AP range by 5.7%. The uni-directional CIs for T1-T2 for AP SD and AP range, tentatively suggest that the two different textured surfaces may have opposite effects on AP sway parameters.

Table 5.2: Postural sway for each textured condition during quiet standing with eyes open over 30 seconds in healthy young adults (n=24)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p=*	C-T1 Mean diff. (95%CI) *p=*	C-T1 % Change	C-T2 Mean diff. (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
AP SD (mm)	4.4 (1.4)	4.1 (0.9)	4.6 (0.8)	0.105	-0.3 (-0.8 to 0.2) *0.178*	-6.8	0.2 (-0.4 to 0.8) *0.509*	4.5	0.5 (0.1 to 0.9) *0.015*
AP range (mm)	24.5 (5.7)	23.1 (3.6)	24.8 (3.0)	0.216	-1.4 (-3.7 to 0.9) *0.232*	-5.7	0.3 (-1.9 to 2.5) *0.776*	1.2	1.7 (0.03 to 3.4) *0.047*
ML SD (mm)	3.0 (0.7)	3.1 (0.7)	3.0 (0.6)	0.669	0.1 (-0.2 to 0.4) *0.445*	3.3	0.01 (-0.3 to 0.3) *0.927*	0.3	-0.1 (-0.3 to 0.1) *0.343*
ML range (mm)	18.2 (3.4)	18.2 (3.3)	18.3 (2.9)	0.957 gg	0.03 (-1.2 to 1.3) *0.958*	0.2	0.1 (-1.3 to 1.5) *0.866*	0.5	0.1 (-0.7 to 0.9) *0.841*
CoP velocity (mm s^{-1})	9.2 (3.2)	10.4 (5.5)	10.8 (5.0)	0.475	1.2 (-1.5 to 3.9) *0.361*	13.0	1.7 (-0.8 to 4.2) *0.182*	18.5	0.5 (-3.0 to 3.9) *0.787*

gg = Greenhouse-Geisser correction for sphericity

5.3.2: Postural sway over 30 seconds with eyes closed in healthy young adults

Table 5.3 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during quiet standing over 30 seconds with eyes closed. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CI for C-T1 for CoP velocity is less conclusive. This CI showed a marked asymmetry (zero close to the upper limit) in the direction of a decrease in sway velocity. Whilst this asymmetry does not reach significance, this observation does tentatively suggest T1 may have the capacity to reduce CoP velocity. Relative to baseline, T1 reduced CoP velocity by 12.9%.

Table 5.3: Postural sway for each textured condition during quiet standing with eyes closed over 30 seconds in healthy young adults (n=24)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p*=	C-T1 Mean diff. (95%CI) *p*=	C-T1 % Change	C-T2 Mean diff. (95% CI) *p*=	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p*=
AP SD (mm)	4.8 (1.2)	4.7 (1.0)	4.8 (1.1)	0.749	-0.1 (-0.6 to 0.3) *0.503*	-2.1	-0.1 (-0.4 to 0.3) *0.688*	-2.1	0.1 (-0.3 to 0.5) *0.685*
AP range (mm)	27.1 (6.0)	27.7 (5.3)	27.8 (4.7)	0.737	0.6 (-1.6 to 2.9) *0.568*	2.2	0.7 (-1.1 to 2.4) *0.458*	2.5	0.01 (-1.8 to 1.8) *0.987*
ML SD (mm)	3.1 (0.6)	3.1 (0.8)	3.2 (0.6)	0.775	0.03 (-0.2 to 0.3) *0.801*	1.0	0.1 (-0.1 to 0.3) *0.407*	3.2	0.1 (-0.2 to 0.3) *0.680*
ML range (mm)	18.8 (3.1)	19.5 (4.3)	19.4 (3.1)	0.566	0.7 (-0.9 to 2.4) *0.362*	3.7	0.6 (-0.8 to 1.9) *0.392*	3.2	-0.2 (-1.6 to 1.3) *0.814*
CoP velocity (mm s^{-1})	13.2 (7.4)	11.5 (4.4)	12.9 (7.4)	0.143	-1.7 (-3.5 to 0.1) *0.068*	-12.9	-0.4 (-2.5 to 1.7) *0.700*	-3.0	1.3 (-0.2 to 2.9) *0.094*

5.3.3: Postural sway over the first 10 seconds with eyes open in healthy young adults

Table 5.4 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during the first 10 seconds of quiet standing with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant. Inspection of the 95% CIs for C-T1 and C-T2 supports that. All contain zero well within the interval.

However, it is worth noting that relative to C, both T1 and T2 appear to increase CoP velocity by 14.8% and 12.0%, respectively.

Table 5.4: Postural sway for each textured condition during quiet standing with eyes open over the first 10 seconds in healthy young adults (n=24)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p*=	C-T1 Mean diff. (95%CI) p=	C-T1 % Change	C-T2 Mean diff. (95% CI) p=	C-T2 % Change	T1-T2 Mean diff. (95% CI) p=
AP SD (mm)	3.4 (0.7)	3.6 (0.9)	3.6 (0.7)	0.606	0.1 (-0.2 to 0.5) *0.397*	2.9	0.2 (-0.2 to 0.6) *0.340*	5.9	0.1 (-0.4 to 0.5) *0.829*
AP range (mm)	17.9 (3.4)	19.0 (3.8)	18.9 (3.1)	0.483	1.1 (-0.7 to 2.9) *0.224*	6.1	0.9 (-1.0 to 2.9) *0.337*	5.0	-0.2 (-2.4 to 2.1) *0.887*
ML SD (mm)	3.0 (0.7)	2.9 (0.6)	2.9 (0.6)	0.576 gg	-0.03 (-0.3 to 0.2) *0.844*	-1.0	-0.1 (-0.4 to 0.2) *0.422*	-3.3	-0.1 (-0.2 to 0.1) *0.282*
ML range (mm)	16.2 (3.1)	15.8 (2.8)	15.8 (2.9)	0.697	-0.5 (-1.8 to 0.8) *0.472*	-3.1	-0.4 (-1.8 to 1.0) *0.547*	-2.5	0.1 (-0.9 to 1.0) *0.903*
CoP velocity (mm s^{-1})	10.8 (3.6)	12.4 (5.7)	12.1 (5.1)	0.468	1.6 (-1.1 to 4.4) *0.229*	14.8	1.3 (-1.1 to 3.8) *0.273*	12.0	-0.3 (-3.7 to 3.1) *0.864*

gg = Greenhouse-Geisser correction for sphericity

5.3.4: Postural sway over the first 10 seconds with eyes closed in healthy young adults

Table 5.5 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during the first 10 seconds of quiet standing with eyes closed. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CI for C-T1 for CoP velocity is less conclusive. This CI showed a marked asymmetry (zero close to the upper limit) in the direction of a decrease in sway velocity. Whilst this asymmetry does not reach significance, this observation does tentatively suggest T1 may have the capacity to reduce CoP velocity. Relative to baseline, T1 reduced CoP velocity by 11.6%.

Table 5.5: Postural sway for each textured condition during quiet standing with eyes closed over the first 10 seconds in healthy young adults (n=24)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p=*	C-T1 Mean diff. (95%CI) *p=*	C-T1 % Change	C-T2 Mean diff. (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
AP SD (mm)	4.3 (1.2)	4.2 (0.8)	4.2 (0.8)	0.697 gg	-0.2 (-0.7 to 0.4) *0.588*	-4.7	-0.1 (-0.5 to 0.2) *0.479*	-2.3	0.03 (-0.4 to 0.5) *0.872*
AP range (mm)	22.2 (5.4)	22.3 (3.6)	22.1 (3.3)	0.974 gg	0.1 (-2.7 to 2.8) *0.963*	0.5	-0.1 (-1.9 to 1.8) *0.930*	-0.5	-0.1 (-1.9 to 1.6) *0.871*
ML SD (mm)	3.1 (0.6)	3.1 (0.8)	3.0 (0.6)	0.845	0.03 (-0.3 to 0.4) *0.858*	1.0	-0.1 (-0.3 to 0.2) 0.690	-3.2	-0.1 (-0.4 to 0.2) *0.549*
ML range (mm)	16.7 (2.9)	17.3 (3.6)	16.7 (3.1)	0.709	0.6 (-1.2 to 2.3) *0.507*	3.6	-0.02 (-1.5 to 1.5) *0.984*	-0.1	-0.6 (-2.2 to 1.1) *0.485*
CoP velocity (mm s^{-1})	16.4 (7.8)	14.5 (4.2)	15.4 (7.6)	0.216	-1.9 (-4.1 to 0.3) *0.090*	-11.6	-1.0 (-3.3 to 1.4) *0.412*	-6.1	0.9 (-1.1 to 3.0) *0.344*

gg = Greenhouse-Geisser correction for sphericity

5.3.5: Postural sway over the latter 20 seconds with eyes open in healthy young adults

Table 5.6 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions over the latter 20 seconds of quiet standing with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CIs for C-T1 and T1-T2 for AP SD are less conclusive. The uni-directional CI for C-T1 does not contain zero and is in the direction of a decrease in sway. The CI for T1-T2 for AP SD also does not contain zero.

The 95% CIs for C-T1 and T1-T2 for AP range are less conclusive. The CI for C-T1 showed a marked asymmetry (zero close to the upper limit) in the direction of a decrease in sway. The CI for T1-T2 for AP range does not contain zero.

Whilst the uni-directional and asymmetrical CIs for C-T1 do not reach significance, these observations tentatively suggest T1 may have the capacity to reduce AP sway relative to C. Standing on T1 brought about a 7.7% reduction in AP SD, and 6.3% reduction in AP range, beyond baseline. Uni-directional CIs for T1-T2 tentatively suggest the two textured surfaces may have opposite effects on AP sway parameters.

Table 5.6: Postural sway for each textured condition during quiet standing with eyes open over the latter 20 seconds in healthy young adults (n=24)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p=*	C-T1 Mean diff. (95%CI) *p=*	C-T 1 % Change	C-T2 Mean diff. (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
AP SD (mm)	3.9 (1.1)	3.6 (0.7)	3.9 (0.7)	0.067	-0.3 (-0.7 to -0.01) *0.043*	-7.7	0.01 (-0.4 to 0.4) *0.966*	0.3	0.4 (0.1 to 0.6) *0.014*
AP range (mm)	20.5 (5.0)	19.2 (2.7)	20.6 (3.2)	0.086	-1.3 (-2.9 to 0.3) *0.096*	-6.3	0.1 (-1.5 to 1.8) *0.855*	0.5	1.5 (0.4 to 2.5) *0.010*
ML SD (mm)	2.7 (0.7)	2.7 (0.6)	2.8 (0.6)	0.828	0.03 (-0.2 to 0.3) *0.837*	1.1	0.1 (-0.2 to 0.3) *0.578*	3.7	0.05 (-0.2 to 0.3) *0.649*
ML range (mm)	14.9 (3.2)	14.7 (2.8)	15.4 (2.8)	0.316	-0.2 (-1.2 to 0.9) *0.741*	-1.3	0.6 (-0.6 to 1.7) *0.313*	4.0	0.7 (-0.2 to 1.7) *0.117*
CoP velocity (mm s^{-1})	8.4 (3.2)	9.4 (5.6)	10.2 (5.3)	0.453	1.0 (-1.8 to 3.8) *0.465*	11.9	1.9 (-0.8 to 4.5) *0.164*	22.6	0.9 (-2.7 to 4.4) *0.621*

5.3.6: Postural sway over the latter 20 seconds with eyes closed in healthy young adults

Table 5.7 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during the latter 20 seconds of quiet standing with eyes closed. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CIs for C-T1 and T1-T2 for CoP velocity are less conclusive. The CI for C-T1 showed a marked asymmetry (zero close to the upper limit) in the direction of a decrease. The CI for T1-T2 for CoP velocity does not contain zero.

Whilst the asymmetrical CI for C-T1 does not reach significance, this observation tentatively suggests T1 may have the capacity to reduce CoP velocity relative to C. Standing on T1 brought about a 12.9% reduction in CoP velocity. The uni-directional CI for T1-T2 tentatively suggests the two textured surfaces may have opposite effects on CoP velocity.

Table 5.7: Postural sway for each textured condition during quiet standing with eyes closed over the latter 20 seconds in healthy young adults (n=24)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p*=	C-T1 Mean diff. (95%CI) p=	C-T1 % Change	C-T2 Mean diff. (95% CI) p=	C-T2 % Change	T1-T2 Mean diff. (95% CI) p=
AP SD (mm)	4.1 (0.9)	4.1 (1.2)	4.2 (1.0)	0.762	-0.02 (-0.4 to 0.3) *0.897*	-0.5	0.1 (-0.2 to 0.4) *0.510*	2.4	0.1 (-0.3 to 0.5) *0.540*
AP range (mm)	21.5 (4.4)	22.1 (5.8)	22.9 (4.6)	0.255	0.5 (-1.2 to 2.3) *0.534*	2.3	1.3 (-0.2 to 2.9) *0.091*	6.0	0.8 (-0.8 to 2.4) *0.318*
ML SD (mm)	2.7 (0.6)	2.8 (0.9)	2.8 (0.7)	0.673	0.03 (-0.2 to 0.3) *0.847*	1.1	0.1 (-0.1 to 0.3) *0.347*	3.7	0.1 (-0.2 to 0.3) *0.521*
ML range (mm)	15.4 (2.8)	15.6 (4.4)	16.1 (3.5)	0.505	0.1 (-1.2 to 1.5) *0.826*	0.6	0.7 (-0.5 to 1.8) *0.252*	4.5	0.5 (-0.7 to 1.7) *0.379*
CoP velocity ($mm\ s^{-1}$)	11.6 (7.4)	10.1 (4.7)	11.6 (7.4)	0.150	-1.5 (-3.3 to 0.3) *0.094*	-12.9	-0.01 (-2.1 to 2.1) *0.995*	-0.1	1.5 (0.1 to 2.9) *0.039*

5.4: EMG during Quiet Standing in Healthy Young Adults

5.4.1: EMG activity over 30 seconds with eyes open in healthy young adults

Table 5.8 shows the pairwise comparisons, magnitude and direction of change in lower limb EMG activity between conditions during quiet standing over 30 seconds with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant. Inspection of the 95% CIs for C-T1 and C-T2 supports that. All contain zero well within the interval.

EMG data collected during the first 10 and latter 20 seconds of quiet standing with eyes open were not explored. As justified in chapter 4 of this thesis, no further EMG analysis was conducted on the basis that no statistically significant differences had been reported between conditions, for any postural sway variables of interest over these same split time intervals.

Table 5.8: Lower limb EMG activity (μV) for each textured condition during quiet standing with eyes open over 30 seconds in healthy young adults (n=24)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p*=	C-T1 Mean diff. (95%CI) *p*=	C-T1 % Change	C-T2 Mean diff. (95% CI) *p*=	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p*=
Vastus Medialis	12.3 (6.1)	12.4 (6.0)	12.8 (6.8)	0.419 gg	0.1 (-0.7 to 0.8) *0.852*	0.8	0.5 (-0.7 to 1.6) *0.421*	4.1	0.4 (-0.2 to 0.9) *0.156*
Rectus Femoris	3.4 (2.3)	3.4 (2.2)	3.6 (2.5)	0.402	0.003 (-0.3 to 0.3) *0.986*	0.9	0.2 (-0.1 to 0.5) *0.264*	5.9	0.2 (-0.1 to 0.4) *0.140*
Vastus Lateralis	6.4 (9.0)	6.7 (8.7)	7.6 (10.8)	0.338 gg	0.3 (-1.4 to 1.9) *0.740*	4.7	1.2 (-1.1 to 3.3) *0.296*	18.8	0.9 (-0.3 to 2.0) *0.127*
Biceps Femoris	3.6 (1.8)	4.2 (3.3)	4.5 (3.7)	0.086 gg	0.7 (-0.2 to 1.5) *0.132*	19.4	1.0 (-1.0 to 2.0) *0.072*	27.8	0.3 (-0.2 to 0.8) *0.197*
Tibialis Anterior	3.6 (3.1)	2.9 (0.8)	3.3 (2.8)	0.267 gg	-0.7 (-1.8 to 0.4) *0.213*	-19.4	-0.3 (-0.6 to 0.04) *0.085*	-8.3	0.4 (-0.6 to 1.4) *0.405*
Peroneus Longus	11.0 (5.5)	10.9 (5.4)	10.6 (5.6)	0.468	-0.1 (-0.9 to 0.7) *0.750*	-0.9	-0.4 (-1.1 to 0.2) *0.202*	-3.6	-0.3 (-1.0 to 0.4) *0.394*
Medial Gastrocnemius	10.0 (9.2)	9.7 (7.0)	10.2 (8.9)	0.736 gg	-0.3 (-2.1 to 1.5) *0.741*	-3.0	0.2 (-0.7 to 1.0) *0.678*	2.0	0.5 (-1.4 to 2.3) *0.616*
Soleus	17.8 (9.8)	17.5 (8.7)	16.9 (8.6)	0.276 gg	-0.4 (-1.9 to 1.1) *0.604*	-2.2	-1.0 (-2.3 to 0.4) *0.158*	-5.6	-0.6 (-1.3 to 0.2) *0.122*

gg = Greenhouse-Geisser correction for sphericity

5.4.2: EMG activity over 30 seconds with eyes closed in healthy young adults

Table 5.9 shows the pairwise comparisons, magnitude and direction of change in lower limb EMG activity between conditions during quiet standing over 30 seconds with eyes closed. The repeated measures ANOVAs showed these differences were not statistically significant. Inspection of the 95% CIs for C-T1 and C-T2 supports that. All contain zero well within the interval.

EMG data collected during the first 10 and latter 20 seconds of quiet standing with eyes closed were not explored. As justified in chapter 4 of this thesis, no further EMG analysis was conducted on the basis that no statistically significant differences had been reported between conditions, for any postural sway variables of interest over these same split time intervals.

Table 5.9: Lower limb EMG activity (μV) for each textured condition during quiet standing with eyes closed over 30 seconds in healthy young adults (n=24)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p=*	C-T1 Mean diff. (95%CI) *p=*	C-T1 % Change	C-T2 Mean diff. (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
Vastus Medialis	13.3 (7.0)	12.8 (7.0)	13.3 (6.9)	0.577	-0.5 (-1.6 to 0.6) *0.330*	-3.8	-0.1 (-1.3 to 1.2) *0.920*	-0.8	0.5 (-0.5 to 1.4) *0.347*
Rectus Femoris	4.1 (3.0)	3.6 (2.7)	4.0 (2.6)	0.251 gg	-0.5 (-1.2 to 0.2) *0.161*	-12.2	-0.1 (-0.8 to 0.5) *0.680*	-2.4	0.3 (-0.1 to 0.7) *0.087*
Vastus Lateralis	7.9 (11.4)	7.8 (11.7)	8.2 (11.4)	0.812	-0.1 (-1.1 to 0.9) *0.807*	-1.3	0.2 (-0.9 to 1.4) *0.679*	2.5	0.4 (-0.9 to 1.6) *0.570*
Biceps Femoris	4.5 (3.5)	5.4 (6.1)	4.0 (2.7)	0.187 gg	0.8 (-0.7 to 2.4) *0.290*	17.8	-0.5 (-1.5 to 0.4) *0.241*	-11.1	-1.4 (-3.2 to 0.5) *0.138*
Tibialis Anterior	4.0 (3.5)	3.4 (2.1)	3.8 (3.7)	0.512 gg	-0.6 (-1.9 to 0.7) *0.346*	-15.0	-0.2 (-1.6 to 1.1) *0.736*	-5.0	0.4 (-0.4 to 1.1) *0.302*
Peroneus Longus	11.1 (5.4)	11.5 (5.4)	11.5 (5.1)	0.463	0.4 (-0.4 to 1.3) *0.316*	3.6	0.4 (-0.4 to 1.1) *0.345*	3.6	-0.1 (-0.7 to 0.5) *0.823*
Medial Gastrocnemius	11.6 (8.6)	11.7 (8.4)	10.5 (7.1)	0.235 gg	0.1 (-0.9 to 1.0) *0.857*	1.9	-1.1 (-3.1 to 0.9) *0.266*	-9.5	-1.2 (-2.8 to 0.4) *0.142*
Soleus	17.6 (8.7)	18.4 (8.9)	18.1 (8.7)	0.354	0.9 (-0.4 to 2.2) *0.184*	5.1	0.5 (-0.7 to 1.8) *0.399*	2.8	-0.3 (-1.5 to 0.8) *0.540*

gg = Greenhouse-Geisser correction for sphericity

5.5: Reliability of Postural Sway Parameters in Healthy Young Adults

5.5.1: Within-session reliability of postural sway parameters in healthy young adults

The results for within-session reliability of postural sway parameters over three repeated trials of quiet standing on the control surface with eyes open over 30 seconds, in healthy young adults are reported in Table 5.10. All postural sway parameters showed moderate (AP SD, AP range, ML SD, ML range) or substantial (CoP Velocity) relative reliability. SEM values for within-session reliability of postural sway measures can also be seen in Table 5.10. As expected, it appears that the magnitude of measurement error increases as the magnitude of mean sway increases.

Table 5.10: Within-session reliability for postural sway variables for three repetitions in healthy young adults (n=24)

Postural Sway Variable	Repetition 1 Mean (SD)	Repetition 2 Mean (SD)	Repetition 3 Mean (SD)	ICCs (95% CI)	SEM
AP SD (mm)	4.4 (1.7)	4.5 (2.0)	4.3 (1.5)	0.67 (0.4 to 0.8)	1.34
AP Range (mm)	23.8 (8.2)	25.5 (8.3)	24.3 (5.5)	0.65 (0.3 to 0.8)	5.84
ML SD (mm)	3.0 (0.9)	3.0 (0.7)	3.1 (0.9)	0.76 (0.5 to 0.9)	0.58
ML Range (mm)	17.6 (4.1)	17.8 (3.5)	19.1 (5.5)	0.66 (0.3 to 0.8)	3.45
CoP Velocity (mm s^{-1})	9.1 (3.2)	8.6 (2.3)	9.8 (5.0)	0.84 (0.7 to 0.9)	2.23

5.6: Reliability of Lower Limb EMG Amplitude in Healthy Young Adults

5.6.1: Within-session reliability of lower limb EMG amplitude in healthy young adults

The results for within-session reliability of lower limb EMG amplitude measurements are presented in Table 5.11. The ICCs are reported to be substantial for VM (0.99), RF (0.99), VL (0.98), PL (0.97), MG (0.97) and SOL (0.98), and good for BF (0.68). As the EMG electrodes remained in place between all repetitions, excellent reliability would be expected for all muscles. However, this was not the case for TA, showing an ICC of 0.22.

BF and TA show high intra-session measurement error of 1.80μV and 4.75μV, respectively. The magnitude of measurement error for BF is close to 50% of its mean amplitude. The SEM for TA exceeds the mean activity observed in this muscle during each of the three repeated quiet standing tests. The remaining muscles, VM, RF, VL, PL, MG and SOL all show some degree of measurement error: however, relative to their mean amplitude scores this error is not as high as that observed for BF and TA.

Table 5.11: Within-session reliability for lower limb EMG Activity (µV) for three repetitions in healthy young adults (n=24)

Muscle	Repetition 1 Mean (SD)	Repetition 2 Mean (SD)	Repetition 3 Mean (SD)	ICCs (95% CI)	SEM
Vastus Medialis	12.5 (6.4)	12.1 (6.0)	12.3 (6.0)	0.99 (0.98 to 1.0)	0.94
Rectus Femoris	3.5 (2.2)	3.2 (2.2)	3.6 (2.4)	0.99 (0.97 to 1.0)	0.47
Vastus Lateralis	6.6 (9.8)	6.2 (9.6)	6.5 (8.0)	0.98 (0.97 to 1.0)	1.98
Biceps Femoris	4.1 (3.2)	3.6 (2.0)	3.1 (1.5)	0.68 (0.4 to 0.9)	1.80
Tibialis Anterior	3.1 (1.3)	3.0 (1.0)	4.7 (8.4)	0.22 (-0.5 to 0.6)	4.75
Peroneus Longus	11.2 (5.4)	11.4 (5.7)	10.4 (5.7)	0.97 (0.9 to 1.0)	1.52
Medial Gastrocnemius	10.1 (9.8)	10.5 (10.1)	9.4 (8.3)	0.97 (0.9 to 1.0)	2.69
Soleus	18.8 (10.0)	18.1 (10.6)	16.6 (9.3)	0.98 (0.97 to 1.0)	2.19

5.7: Summary

Healthy young adults were reported to have full sensation on the plantar surface of their feet. Observations of foot type indicated that 91.7% of the study sample had normal foot structure. There were insufficient participants presenting foot pronation (low-arched) or supination (high-arched) to justify sub-group analysis.

The findings from this study, investigating the effect of textured surfaces on postural sway and lower limb muscle activity in healthy young adults, point to a number of suggestions. The evidence presented in this chapter tentatively suggests textured surfaces may have the capacity to alter AP sway parameters and CoP velocity in healthy young adults.

During quiet standing with eyes open, asymmetrical and uni-directional CIs for C-T1 tentatively suggest standing on T1 can reduce AP SD and AP range, beyond baseline. During quiet standing with eye closed, asymmetrical CIs for C-T1 suggest standing on T1 can reduce CoP velocity, beyond baseline. Uni-directional 95% CIs for T1-T2, suggest the two different textured surfaces may have opposite effects on AP sway and CoP velocity.

Textured surfaces were reported to have no significant effects on the amplitude of lower limb EMG activity in any of the 8 muscles of interest during quiet standing with eyes open or eyes closed.

This chapter shows that all postural sway parameters have good or substantial reliability during short-term standing balance tests. Average EMG amplitude is also a reliable parameter for measuring muscle activity in VM, RF, VL, BF, MG, PL and SOL during repeated tests of quiet standing. TA was shown to have poor within-session reliability. Therefore caution must be taken when interpreting the findings for the effect of texture surfaces on the average amplitude of TA activity.

CHAPTER 6: DISCUSSION OF THE EFFECT OF TEXTURED SURFACES ON POSTURAL SWAY AND LOWER LIMB MUSCLE ACTIVITY DURING QUIET STANDING IN HEALTHY YOUNG ADULTS

6.1: Introduction

This chapter will discuss the experimental findings for healthy young adults, within the context of previous literature relating to postural control, muscular contributions to upright standing, and the use of texture as a medium to enhance balance performance in healthy young adults. The chapter will conclude discussing the findings from the reliability study in healthy young adults.

6.2: Textured Surfaces and Postural Sway in Healthy Young Adults

The current study reported that two textured surfaces, which differed only in their pattern of indentation, did not significantly affect AP or ML sway or CoP velocity in comparison to a smooth, control surface during quiet standing in healthy young adults (Hatton et al., 2009) (Appendix 14). No statistically significant findings were reported when standing quietly with eyes open and eyes closed, over the first 10, latter 20 and overall 30 second time intervals (Tables 5.2 to 5.7). However, as will be discussed in this chapter, these findings should not be interpreted as conclusive. The distribution of scores within the 95% CI for pairwise comparisons, tentatively suggest textured surfaces may have the capacity to alter AP sway and CoP velocity in healthy young adults.

Baseline postural sway data observed in the current study for quiet standing over 30 seconds with eyes open, resembles the magnitude of sway reported in previous literature, during similar test conditions (Benjuya et al., 2004, Laughton et al., 2003, Raymakers et al., 2005) (Table 6.1).

Table 6.1: Comparative baseline postural sway data during quiet standing with eyes open on a flat, rigid surface in healthy young adults between the current and previous studies

Study	Participants	Test Conditions	AP Sway (mm) Mean (SD)	ML Sway (mm) Mean (SD)	CoP Velocity (mm s^{-1}) Mean (SD)
Current study	Healthy young adults	Quiet standing, 30 seconds, eyes open	*APSD :* 4.4 (1.4) *AP Range :* 24.5 (5.7)	*MLSD :* 3.0 (0.7) *ML Range:* 18.2 (3.4)	*CoP Velocity :* 9.2 (3.2)
Laughton et al. (2003)	Healthy young adults	Quiet standing, 30 seconds, eyes open	*APSD:* 3.5 (1.0) *AP Range:* 16.7 (5.0)	*MLSD:* 2.3 (0.8) *ML Range:* 10.9 (3.2)	-
Benjuya et al. (2004)	Healthy young adults	Quiet standing, 20 seconds, eyes open, feet together	*AP sway:* 15.0 (1.0)	*ML sway:* 20.0 (1.0)	*CoP Velocity:* 12.0 (0.4)
Raymakers et al. (2005)	Healthy young adults	Quiet standing, 60 seconds, eyes open	*AP Range:* 23.0 (7.0)	*ML Range:* 18.0 (5.0)	*CoP Velocity:* 9.4 (1.9)

The results of the current study are in agreement with those of Wilson et al. (2008). The textured interventions investigated in both these studies were exactly the same materials, and thus had the same geometric patterns and Shore Value. T1 was referred to as the grid insole by Wilson et al. (2008), whilst T2 was labelled the dimple insole. However, there are a number of dissimilarities in methodology between these two studies including the mode of textured intervention, whether a foot insole or surface, the inclusion of hosiery and standardised footwear, the age and gender of study participants. Irrespective of their different test procedures, collectively the results from the current study and that of Wilson et al. (2008) suggest using texture as a medium to augment plantar tactile stimulation and enhance balance performance, may not be effective in healthy young or middle-aged adults. Such conclusions are based on the absence of any statistically significant changes in balance performance between textured conditions.

However, Wilson et al. (2008) speculated that their textured insoles were not detrimental to balance performance. It is possible that any significant, beneficial textured effect may have been masked by methodological limitations. Similarly, the current study tentatively reported that textured surfaces may have the capacity to alter AP sway and CoP velocity during quiet standing in healthy young adults. These observations may have failed to reach the level of statistical significance due to the methods of statistical analysis used in the current study. Some statisticians recommend using 90% CI, rather than the 'standard' 95% CI. Whilst the alpha level is commonly set at 5% throughout literature, this value is quite arbitrary (Hopkins et al., 2009).

In the current study, speculative observations for the effect of textured surfaces on CoP velocity may support the findings of Corbin et al. (2007) who reported that textured insoles have some beneficial interactions on CoP velocity. As discussed in chapter 2 of this thesis, Corbin et al. (2007) concluded that textured insoles showed a significant interaction between visual condition and texture during quiet bilateral standing. In this context, the term 'interaction' is understood to mean that the effect of one independent variable (textured insole) on the dependent variable (postural sway) depends on a second independent variable (visual condition). When wearing textured insoles, there was no significant difference in CoP area and CoP velocity between eyes open versus eyes closed conditions (Corbin et al., 2007). Throughout literature there is general consensus that balance performance declines when visual sensory information is removed (Goldie et al., 1992, Hasan et al., 1990, Shumway-

Cook and Horak, 1986, Winter, 1995). Therefore, Corbin et al. (2007) concluded that enhanced tactile plantar stimulation, provided by the textured insoles, may provide a surrogate source of sensory information which is capable of counteracting the effect of deprived visual inputs on postural sway measures. In the current study, the statistical interaction between visual and textured conditions was not formally analysed. However, the results of the current study suggest that T1 may have the capacity to reduce CoP velocity when standing with eyes closed. Therefore, both Corbin et al. (2007) and the current study observed some interaction of texture on measures of CoP velocity, irrespective of the level of statistical significance. This agreement in findings may be attributed to the sensitivity of CoP velocity, which is considered to be one of the most reliable sway measures for determining postural stability (Lafond et al., 2004, Lin et al., 2008, Raymakers et al., 2005, Takala et al., 1997). This finding is important from both a clinical and sports perspective. When the CoP moves more slowly, the CoP takes a longer duration to reach the limits of stability. This provides greater time for an individual to execute corrective postural responses to internal or external perturbations. This can be applied clinically where balance impairment originates from injury or pathology or to a sporting environment where sudden cutting movements can be de-stabilising. If a textured intervention can be used to significantly slow down the rate at which the CoP moves, then such an intervention may ‘buy’ an individual extra time to correct their posture and maintain upright standing balance. This could prevent a fall or improve sports performance.

There are a number of methodological dissimilarities between the current study and that of Corbin et al. (2007), which may explain differences in the significance of the textured effect on sway parameters. Firstly, the textured intervention must be considered. The geometric pattern of textured surfaces investigated in the current study appear to be less prominent than that of the textured insole used by Corbin et al. (2007), and may therefore indent the skin to a lesser degree. Rounded plastic nubs on the textured insole were raised approximately 2.5mm from the surface (Corbin et al., 2007) and appear to be sharper and greater in height (Fig 6.1) than indentations on the textured surfaces displayed in chapter 4 (Fig 4.6) of this thesis. Centre-to-centre distances between protrusions on the textured insole, also appear to be greater than that of the textured surfaces. This is important as chapter 2 of the current thesis highlights that plantar mechanoreceptors respond to stimuli such as indenting or stretching of the skin through contact with a surface (Sherwood, 1997). Therefore, a more prominent textured pattern may lead to greater plantar stimulation and account for the different effects of texture on postural sway observed in the current study, compared to Corbin et al. (2007).

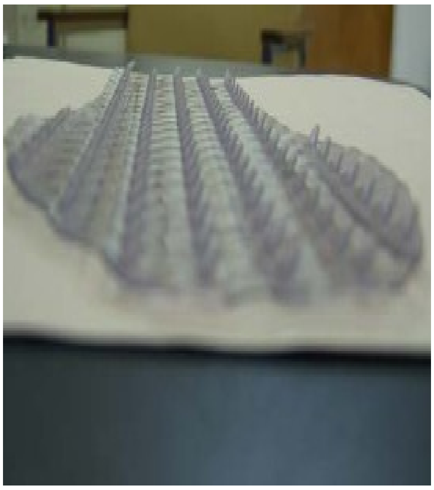

Fig 6.1: Textured insole investigated by Corbin et al. (2007)

Corbin et al. (2007) investigated the effect of textured insoles, worn within shoes, on balance performance. The current study explored the effect of textured surfaces whilst standing barefoot. Considering the significant findings, reported by Corbin et al. (2007), it is possible that their use of footwear may have served to hold the textured insole more closely to the plantar surface of the foot. This may have enhanced contact between the protrusions and cutaneous mechanoreceptors, leading to greater tactile stimulation. However, in the study by Corbin et al. (2007) all participants wore thin cotton socks. It is possible that wearing hosiery could essentially dampen the textured effect. This did not appear to be the case, although the effect of textured insoles on CoP area and velocity may have been greater should participants have been barefoot.

In the current study, all participants remained barefoot when standing on the textured surfaces, to optimise contact between protrusions and the foot sole. The effects of textured surfaces on sway parameters did not reach significance. This evidence points to the suggestion that differences in findings between Corbin et al. (2007) and the current study could be attributed to the textured pattern, reinforcing postulations that there may be an optimal geometric design for enhancing balance performance. It is possible that wearing cotton socks resembled participants' normal daily life, thus the only alteration was the addition of the textured sensory stimulus (Corbin et al., 2007). In the current study participants stood barefoot: this may have caused thermal changes in the skin, and exposure to a new source of tactile sensory stimulus to which the body may not have been able to adapt.

Although the temperature of the laboratory was kept constant at approximately 22ºC, it is not customary for Westerners to carry out daily activities barefoot or unshod. Exposing participants' barefeet to a cool environment may have led to changes in foot temperature. It is unknown within what temperature range plantar mechanoreceptors optimally function. Therefore, should thermal changes have occurred at the feet, whereby cutaneous temperatures dropped below this optimal range, then normal plantar mechanoreceptor function, may have been impaired. This could explain why any textured effect on balance performance did not reach the level of significance. In the current study, standing barefoot may have brought about similar physiological effects to that reported following induced plantar hypoanaesthesia through cooling (Eils et al., 2004, Eils et al., 2002).

Further differences in test methodology exist between the current study and that of Corbin et al. (2007). In the current study, the control condition was a smooth flat sheet of plain EVA material, which had no indentations. Corbin et al. (2007) explored the effect of a textured insole on sway parameters compared to a control condition, where participants wore nothing in their shoes. This raises issues relating to in-shoe volume between texture and no texture conditions, and the possibility that significant differences in sway parameters may be attributable to inconsistencies in foot pressure or comfort.

Corbin et al. (2007) collected sway data over 10 seconds. Sway data collected over this time interval may be highly variable due to stabilisation of participants (Le Clair and Riach, 1996) or the force platform (Raymakers et al., 2005), or alternatively represent short-term exploratory activity of the CNS (Mochizuki et al., 2006). If exploratory activity takes place for each different textured surface, so the CNS can establish the limits of stability for every new textured pattern, then it may be unlikely that any true therapeutic effect of texture could be detected within this time interval. Thus, the significant interaction between texture and visual condition, observed by Corbin et al. (2007), may represent large variability in sway measures, rather than a true textured effect. The current study took these confounding factors into consideration, collecting sway data over the first 10, latter 20 and overall 30 seconds of quiet standing.

In the current study, the tentative finding for textured surfaces to have opposite effects on AP sway appears to be in agreement with significant findings from Palluel et al. (2008). Palluel et al. (2008) concluded that plantar tactile stimulation significantly enhances balance performance in healthy young adults, having observed improvements in AP and ML sway parameters during quiet standing, after wearing spike insoles for 5 minutes. Textured ‘spike’ insoles had no significant effects on postural sway at the point of application, suggesting plantar cutaneous mechanoreceptors may require a period of adaptation to new sources of sensory stimuli. It is possible that textured surfaces or insoles may provide altered sensory information that the CNS cannot use immediately or in the short-term, for balance control. This altered sensory information may constrain the sensorimotor system, thereafter leading to no observable changes in balance performance. This may explain why the effect of textured surfaces on postural sway in healthy young adults during quiet standing over short durations in the current study did not reach significance, and immediately after putting on textured insoles in the study by Palluel et al. (2008). It is possible that there may be an optimal exposure time to textured footwear interventions, in order to observe significant alterations in bilateral standing balance, which previous studies have not yet identified (Corbin et al., 2007, Palluel et al., 2008, Wilson et al., 2008). Under-exposure to texture may provide insufficient time for plantar mechanoreceptor adaptation, whilst over-exposure may render a textured footwear intervention ineffective.

There are a number of limitations in the methodology and reporting of results, by Palluel et al. (2008). This makes it difficult to evaluate the clinical implications of their textured insole on balance performance and explain conflicting findings with the current study.

The textured pattern, specific to the spike insoles, comprised uniformly distributed indentations over the full plantar aspect of the foot (Palluel et al., 2008). However, ‘spikes’ located under the medial arch of the foot were of greater height, diameter and density than spikes contacting fore- and rearfoot regions. Therefore, it is possible that larger protrusions in the mid-foot region essentially provided a contouring effect, ensuring all regions of the plantar surface of the foot received a similar degree of stimulation. In the current study, the textured surfaces were sheets of soling material, which were not contoured to participants’ feet. As the height of the medial arch differs between individuals, this suggests that the degree of tactile stimuli received by mid-foot plantar mechanoreceptors could have varied throughout the study sample, according to participants' foot posture. It is possible that insufficient stimulation to mid-foot regions may explain why the current study did not observe comparable findings to Palluel et al. (2008). However, research suggests mechanoreceptors located in the mid-foot region are used to a lesser extent than fore- and rear-foot mechanoreceptors, for balance control (Chiang and Wu, 1997, Meyer et al., 2004).

In the current study, foot posture was assessed for all participants using the FPI6. The purpose of conducting this screening test was to identify the proportion of participants who had normal, low-arched or high-arched foot structure, in order to explore whether contact area between the foot sole and textured surface was a factor which influenced the magnitude of the textured effect. However, as 91.7% of the study sample were reported to have normal foot structure there were insufficient participants within the low- and high-arched groups to support this analysis.

Similar to Corbin et al. (2007) there is a lack of experimental control between baseline and textured conditions (Palluel et al., 2008). In the study by Palluel et al. (2008), the experimental condition involved wearing a pair of spike sandals. The control condition involved wearing these same spike sandals, but with a smooth 3mm foot insole inserted to prevent cutaneous contact with the spikes. Whilst these authors suggest the addition of the control insole did not affect the flexibility of the sandal (Palluel et al., 2008), inserting an additional insole would essentially reduce in-shoe volume. Therefore significant alterations in balance performance between control and spike conditions may be due to changes in foot pressure or comfort, rather than enhanced tactile stimulation.

The results from Palluel et al. (2008) are poorly presented and difficult to interpret. Percentage changes in balance performance between texture and no texture conditions are reported, with positive percentage changes representing improvements in postural stability. This is conflicting, as decreased CoP movement suggests better balance performance. Mean (SD) sway data is displayed in graphs, but it is difficult to interpret actual scores, the true effect size of the textured insoles, and thus their clinical significance.

The current study suggests textured surfaces may be capable of influencing AP sway and CoP velocity parameters in healthy young adults, during quiet standing. ML sway parameters remained unchanged. The two different textured surfaces had a tendency to bring about opposite effects on AP sway when standing with eyes open. Relative to control, T1 reduced, and T2 increased AP sway. Thus, it would appear that T1 may be of greater therapeutic benefit to balance performance in healthy young adults. These trends were not observed during the first 10 seconds of quiet standing, suggesting that the importance of an ‘optimal’ geometric pattern for enhancing balance performance may only become apparent after an initial period of re-stabilisation. These findings indicate a possible effect of texture type on postural sway parameters. However, it remains unclear whether a more prominent textured pattern would elicit greater changes in postural sway measures.

Previous research has established that AP sway is of higher magnitude than ML sway in healthy young adults (Mochizuki et al., 2006). The current study supports this evidence as the magnitude of AP sway always exceeded that of ML sway, irrespective of visual condition or time interval. Therefore, it is possible that textured surfaces may only influence sway parameters in the direction of greatest magnitude, under any given test condition.

A recent review of the literature (Hatton et al., 2008) (Appendix 1), indicates there are no other studies which have investigated the effect of textured surfaces on postural sway and lower limb muscle activity, during quiet standing in healthy young adults. To date, only one other research group has investigated the effect of textured foot insoles on sensorimotor function (Waddington and Adams, 2000, Waddington and Adams, 2003). In two separate studies, it has been shown that textured insoles can significantly alter ankle movement discrimination in young adults (Waddington and Adams, 2000, Waddington and Adams, 2003). Throughout current literature, the unanimous underlying principle for the use of texture is to enhance plantar sensory information (Corbin et al., 2007, Palluel et al., 2008, Waddington and Adams, 2000, Waddington and Adams, 2003). However, the current study is novel, in that it investigated two different patterns of indentation (one texture comprising a small pyramidal design and the other a small circular convex pattern) with the aim of providing a different quality or quantity of sensory input.

The clinical implications of the current findings suggest that textured surfaces are capable of altering AP sway and CoP velocity in healthy young adults. Textured surfaces may be beneficial when used as part of a rehabilitation programme for young adults with balance deficits, pathology or post-injury. However, textured surfaces or footwear interventions should not be used to enhance balance ability in isolation. Concurrent therapeutic interventions, such as exercise, should focus on maximising the textured effect by further improving AP postural control, whilst also focussing on ML sway: an aspect of balance which textured surfaces appear to have no effect in young adults.

6.3: Textured Surfaces and Lower Limb EMG Activity in Healthy Young Adults

The findings from the current study concluded that textured surfaces do not significantly alter lower limb muscle activity during bipedal quiet standing in healthy young adults. This is in conflict with previous research which has shown that during dynamic test conditions, textured footwear interventions significantly affect lower limb muscle activity in healthy young adults (Nurse et al., 2005) and those with Multiple Sclerosis (Kelleher et al., 2010). Textured insoles were observed to decrease overall EMG activity of soleus and tibialis anterior for the entire stance phase during gait trials (Nurse et al., 2005). In comparison, increased activity in medial and lateral gastrocnemius was observed during gait trials in adults with Multiple Sclerosis (Kelleher et al., 2010). These are the only two studies to date, showing texture to significantly influence lower limb muscle activation patterns (Kelleher et al., 2010, Nurse et al., 2005); with neither providing evidence for the effect of texture on EMG activity during static balance tests.

Postural stability during balance tasks can be improved by increasing lower limb muscle activity which increases the rigidity of the lower limb joints (Accornero et al., 1997). However, if the balance task is minimally demanding, such as quiet bilateral standing, then this muscular control mechanism may not need to be activated. This could explain why textured surfaces had no effect on muscle activity in the current study. It is also worth noting that calculating basic average EMG activity over a 30 second time interval may not be sensitive enough to detect any significant changes in muscle activity between textured conditions.

Alternatively, changes in EMG activity may have occurred in other muscles which were not explored, including trunk muscles providing core stability and intrinsic muscles of the toes and feet which grasp a supporting surface. In particular, toe flexor muscle strength can play an important role in controlling postural sway, particularly AP movement, during quiet standing. The stronger the toe flexor muscles, the greater their capacity for grasping the supporting surface, thus enhancing balance performance (Menz et al., 2006). Plantarflexor and toe flexor strength are considered strong determinants of the limits of the functional base of support (Endo et al., 2002). This means that the stronger the foot and ankle muscles, the more anteriorly the ground reaction force can move, relative to the boundary of the base of support (Endo et al., 2002).

It is possible this toe-grasping mechanism may have been put into action when healthy young adults stood on each different textured surface. In the current study, EMG activity in TA could provide some indication of the magnitude of activity in the plantarflexors, which press the feet and toes into the ground. However, no significant differences were reported for TA activity between the textured surfaces for any test condition.

Textured surfaces may provide a medium which is capable of reducing the demand placed on muscles working at high intensities to control upright balance. The results from the current study cannot support this postulation, but provide exploratory data which requires further investigation.

6.4: Discussion of Within-Session Reliability of Postural Sway Parameters in Healthy Young Adults

The current study concluded that when healthy young adults stood quietly with their eyes open for 30 seconds, all postural sway parameters had good or substantial within-session reliability over three repeated trials. This suggests any significant changes in postural sway in healthy young adults may reflect a true textured effect, rather than being representative of unreliable outcome measures. Pinsault and Vuillerme (2009) reported comparable ICCs for AP range (ICC: 0.65), ML range (ICC: 0.67) and CoP velocity (ICC: 0.84) during quiet standing with eyes closed. Together, these findings suggest that irrespective of visual condition, CoP velocity is a substantially reliable sway parameter. This reinforces postulations that CoP velocity is the most consistent and reliable stability parameter (Lafond et al., 2004, Lin et al., 2008, Raymakers et al., 2005, Takala et al., 1997).

However, there is limited previous research, reporting the reliability of postural sway parameters in healthy young adults during quiet standing, to which findings from the current study can be compared. This may be due to the absence of a gold standard procedure for measuring and analysing standing balance.

High ICCs are interpreted to mean that variance observed within a sample can be attributed to differences in between-subject measurements, rather than systematic or random errors. Thus, low ICCs can emerge through two mechanisms: large intra-session variance in outcome measures or minimal between-subject variance. chapter 5 (Table 5.10) of the current thesis reported little variance in mean values of CoP movement over three trials of quiet standing. This may point to two important issues. Firstly, this minimal variation in sway parameters may have contributed to the moderate ICC scores. Secondly, ICCs may lack clinical significance (Liaw et al., 2008).

In the current study, increasing the number of standing trial repetitions may have reduced the error variance compared to the true score, leading to higher ICCs. Pinsault and Vuillerme (2009) reported that substantial within-session reliability for AP range was achieved after 5 consecutive standing tests (ICC: 0.95, 95% CI: 0.73-0.98), 8 repeated tests for ML range (ICC: 0.92, 95% CI: 0.73-0.98) and 2 tests for CoP velocity (ICC: 0.94, 95% CI: 0.83 to 0.98). However, within previous research it is considered acceptable and common practice to perform three repeated trials of standing balance (Hertel et al., 2005, McKeon and Hertel, 2007, Olmsted and Hertel, 2004, Palluel et al., 2008).

In the current study, the SEM results show quite high levels of variability relative to the mean values for all postural sway parameters. However, at a group level, even if the measurement error is substantial, a large enough sample will still permit detection of change with adequate precision (Hopkins, 2000). Furthermore, test procedures including standardised foot, upper body and head positioning, were implemented throughout all repeated trials to minimise potential confounding factors on variability of sway measures. Additionally, the short duration of the testing session was assumed not to provide scope for any physiological diurnal changes, nor lead to fatigue in healthy young adults.

Previous research suggests unperturbed quiet standing tests may be insufficiently demanding for healthy young adults (Brouwer et al., 1998, Moe-Nilssen, 1998). Thus, the simplicity of the balance task used in the current study may be a potential source explaining high measurement error.

6.5: Discussion of Within-Session Reliability of Lower Limb EMG Amplitude in Healthy Young Adults

The current study concluded that when healthy young adults stood quietly with their eyes open for 30 seconds, within-session reliability of EMG amplitude over three repeated trials was reported to be substantial for VM, RF, VL, PL, ML and SOL. Moderate and poor within-session reliability were observed in BF and TA, respectively.

To the author's knowledge, there is no current evidence concerning the within-session reliability of lower limb EMG amplitude measures during quiet standing. Previous research exploring the reliability of lower limb EMG measures, are not directly comparable to the current study, due to differences in EMG parameters, muscles of interest, and contraction-type. Many EMG reliability studies assess maximal and sub-maximal isometric and isokinetic contractions during dynamic or fatiguing activities (Callaghan et al., 2009, Claiborne et al., 2009, Kollmitzer et al., 1999, Laplaud et al., 2006, Mathur et al., 2005, Smoliga et al., 2010).

In the current study, BF was shown to have only moderate within-session reliability, accompanied by high measurement error. One possible explanation is that BF is a bi-articular muscle. Any variation in the degree of knee flexion, or overall biomechanical alignment, between quiet standing trials could lead to inconsistent tension in BF, potentially leading to poor reliability of EMG measures. This may be relevant to the current study as lower limb position was standardised by asking participants to adopt their normal, comfortable standing position, rather than using predetermined criteria such as knee angle. However, this postulation does not appear to hold true for RF and MG, which are also bi-articular muscles and were also investigated.

In the current study, TA was reported to have poor within-session reliability over three repeated tests of quiet standing. TA is a mono-articular muscle, crossing only the ankle joint, and therefore, is not subject to confounding effects of multi-joint movement. Research has shown that alterations in ankle joint position can change TA muscle force (Lee and Piazza, 2008). However, in the current study, foot positioning was standardised using foot templates, in order to minimise variability in the length of the moment arm, and thus levels of lower limb muscle activity.

It is unclear why BF and TA should act differently to the remaining six muscles under investigation, whilst standing on the control surface. It is unlikely that moderate and poor within-session reliability reported in these muscles was attributed to their bi- or mono-articular mechanisms, as this was not the case for other muscles working under similar conditions. Neither can moderate or poor within-session reliability be attributed to alterations in lower limb joint biomechanics as foot positioning was standardised during all test procedures.

It is possible that moderate and poor reliability in BF and TA may have occurred due to large inter-trial variation in AP sway. As already highlighted in this chapter, AP sway is of higher magnitude than ML sway in healthy young adults (Mochizuki et al., 2006). BF activity will arrest excessive anterior sway, whilst TA activity responds to excessive posterior sway. Therefore, large variability in AP sway over repeated trials of quiet standing may contribute to the low ICCs reported in BF and TA. This finding should not be interpreted as a limitation, but may reflect naturally occurring variations in sway patterns and corresponding leg muscle activity in healthy young adults, thus representing normal function.

6.6: Development of Methodology for Healthy Older Adults from Healthy Young Adult Findings

The current study provides speculative evidence that the two different textured surfaces have the capacity to alter AP sway and CoP velocity in healthy young adults. It is possible that a stronger textured effect may be observed in individuals whose baseline balance ability is worse than that of healthy young adults. On that basis, the findings from the healthy young adults supported the original aims of this thesis to explore the effect of texture in a population of healthy older adults. The methodology for healthy older adults was developed upon reflection of findings from healthy young adults.

6.7: Summary

Textured surfaces were shown to have no statistically significant effects on any sway parameter of interest during quiet standing in healthy young adults, irrespective of visual condition or time interval. However, the current study tentatively suggests that the two different textured surfaces may have opposite effects on AP sway in young adults. T1 showed a tendency to reduce, and T2 to increase the magnitude of AP sway parameters. T1 appeared to reduce CoP velocity beyond baseline.

There may be an optimal geometric pattern or textured design, which will bring about changes in sway parameters. It is also possible that a period of adaptation to a textured surface may be required before any functional effects are observed.

Textured surfaces did not change the amplitude of activity in any lower limb muscle of interest during quiet standing in healthy young adults. Textured surfaces may have altered EMG activity in other muscles not investigated in the current study, such as hip or toe musculature. Alternatively, healthy young adults may have successfully used the enhanced sensory information for maintaining balance so that other postural control mechanisms, such as increased lower limb muscle activity, were not required.

Postural sway measures of AP and ML SD, AP and ML range showed moderate within-session reliability. CoP velocity showed substantial reliability. These findings indicate traditional sway parameters are sufficiently reliable for determining the short-term effects of textured surfaces on quiet standing balance.

Average EMG amplitude is a reliable parameter for measuring muscle activity in VM, RF, VL, BF, MG, PL and SOL during repeated tests of quiet standing. TA was observed as being the least reliable muscle. However, it should not be discounted that poor statistical findings may actually represent normal variability and muscle function in healthy young adults.

CHAPTER 7: METHODOLOGY FOR THE EFFECT OF TEXTURED SURFACES ON POSTURAL SWAY AND LOWER LIMB MUSCLE ACTIVITY DURING QUIET STANDING IN HEALTHY OLDER ADULTS

7.1: Introduction

This chapter will present the methods and procedures implemented in healthy older adults. Full details of instrumentation, participant preparation, and test procedures can be found in chapter 4 of this thesis, as they were identical to those in healthy young adults. This chapter will only outline developments to the methodology used in healthy young adults, which were adopted specifically for implementation with healthy older adults to account for age-related declines in physiological and psychological functioning.

7.2: Background

Increasing age leads to deterioration of balance performance through age-related changes in sensorimotor function, which heighten the risk of falling in older adults (Lord et al., 1996). Physiotherapy is seen as having a major role in improving older peoples postural stability (Campbell et al., 1999, Campbell et al., 1997, Gardner et al., 2000), with current techniques including exercise to enhance flexibility and muscle strength around the trunk and lower limbs, coordination, and balance re-training. Whilst exercise programmes may improve balance and help older people regain independence, there is still room to improve the clinical effectiveness of physiotherapy interventions and develop additional methods of reducing falls risk such as the use of simple, non-invasive floor surfaces or footwear interventions.

Previous studies provide a theoretical basis to suggest footwear interventions may help improve balance in older adults. Textured surfaces and foot insoles have been shown to improve static and dynamic balance performance in healthy young adults (Corbin et al., 2007, Palluel et al., 2008, Watanabe and Okubo, 1981). Only one study has reported the effect of textured foot insoles on balance performance in older adults (Palluel et al. 2008). This limited area of research suggests textured interventions may have the capacity to ameliorate age-related declines in balance performance, by optimising sensory input at the feet. Speculative evidence from the first study of this thesis, investigating healthy young adults, points to the possibility of a textured effect in people with poor balance ability: such as older adults.

Previous research demonstrates that by altering the texture of a foot insole, changes can also occur in lower limb muscle activity in healthy young adults (Nurse et al., 2005) and adults with Multiple Sclerosis (Kelleher et al., 2010) during walking. However, it is unknown whether texture affects lower limb muscle activity in older adults, as a component of sensorimotor function of balance control.

This work is novel in that, to date, no other study has explored the effect of different textured surfaces on postural sway and lower limb muscle activity in healthy older adults. It was considered unethical to immediately explore the unknown effects of texture on balance in frail, high-risk older adults. This exploratory study will generate vital evidence relating to the future potential for textured surfaces to improve balance in older people prone to falling.

7.3: Recruitment and Sampling Method

Convenience and snowball sampling methods were used to identify healthy community-dwelling older adults >70 years from the local area for participation. Recruitment posters were located within community organisations and sports clubs (Appendix 15). Older adults recruited via snowball sampling, having spoken with friends or family who were current participants in the study, were sent recruitment letters and information sheets (Appendix 16) through the post.

7.4: Participants, Inclusion and Exclusion Criteria

A cross-sectional sample, comprising 50 healthy older adults (21M, 29F) >70 years, was recruited from Age Concern, Hardwick Baptist Church and local organisations including golf, bowling and leisure clubs. Exclusion criteria were a self-reported history of neuromuscular disease, stroke, peripheral sensory neuropathy, inner ear disorders, inability to walk 10 metres unassisted or inability to stand up and sit down without using their hands. Basic, demographic data including weight, height, BMI, eye height and knee height were measured for all participants. Screening of all older adults involved a short written questionnaire detailing current health and fitness status including any surgery received within the previous 12 months, assessment of foot posture using the FPI6 (Redmond et al., 2006), bilateral foot sensation using monofilaments, and cognition using the MMSE (Folstein et al., 1975) (Appendix 17).

7.5: Electrode Placement and Orientation

In healthy older adults, there were five muscles of interest including RF, VL, BF, TA and MG (Fig 7.1 and 7.2). In comparison to the healthy young adults, data for VM activity was not collected, as greater amounts of sub-cutaneous tissue surrounding the thigh and knee regions in older adults can affect the reliability of electrode placement and EMG signal acquisition. Similarly, PL and SOL were not explored in healthy older adults. Locating the PL and SOL muscle bellies can be difficult in older adults presenting muscular atrophy. Older adults also commonly present with ankle and foot oedema (Dunn et al., 2004). Therefore, increased interstitial lower leg volume may affect the ability to accurately place electrodes over ankle musculature, including SOL.

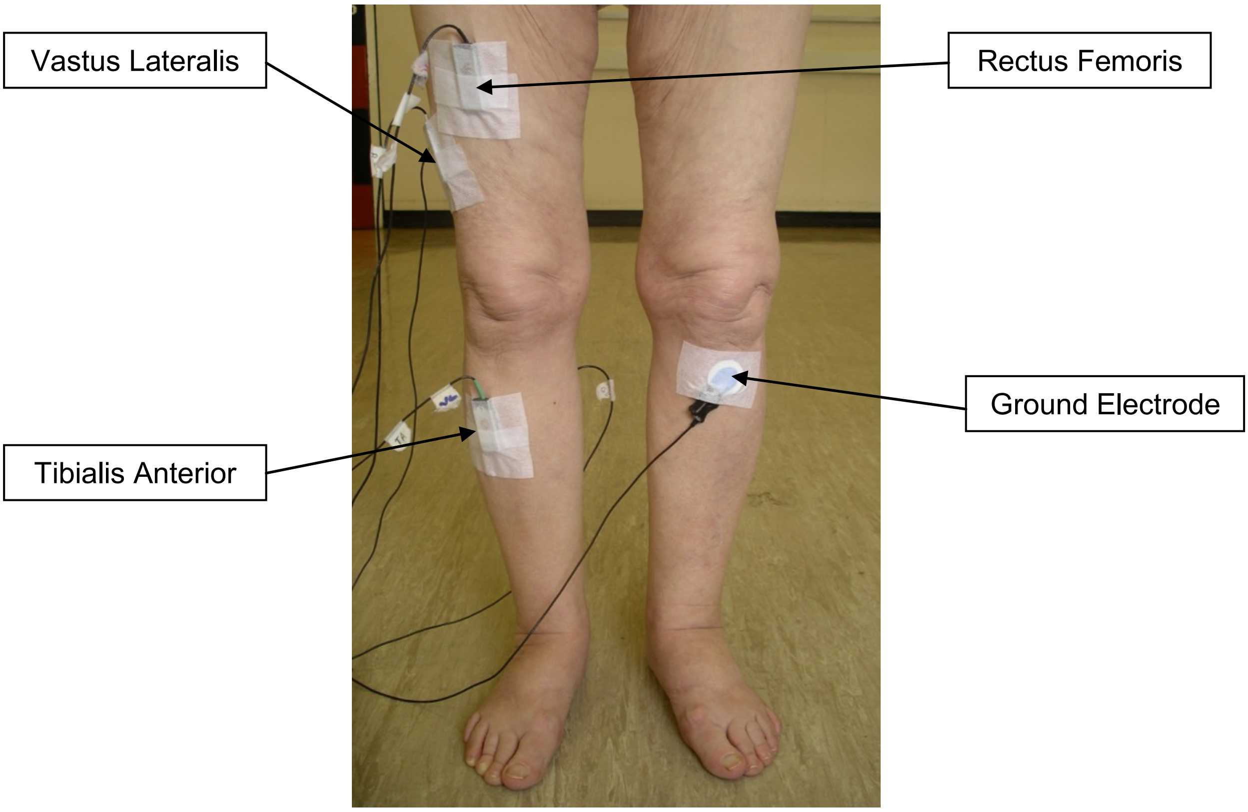

Fig 7.1: EMG electrode placement for healthy older adults – anterior view

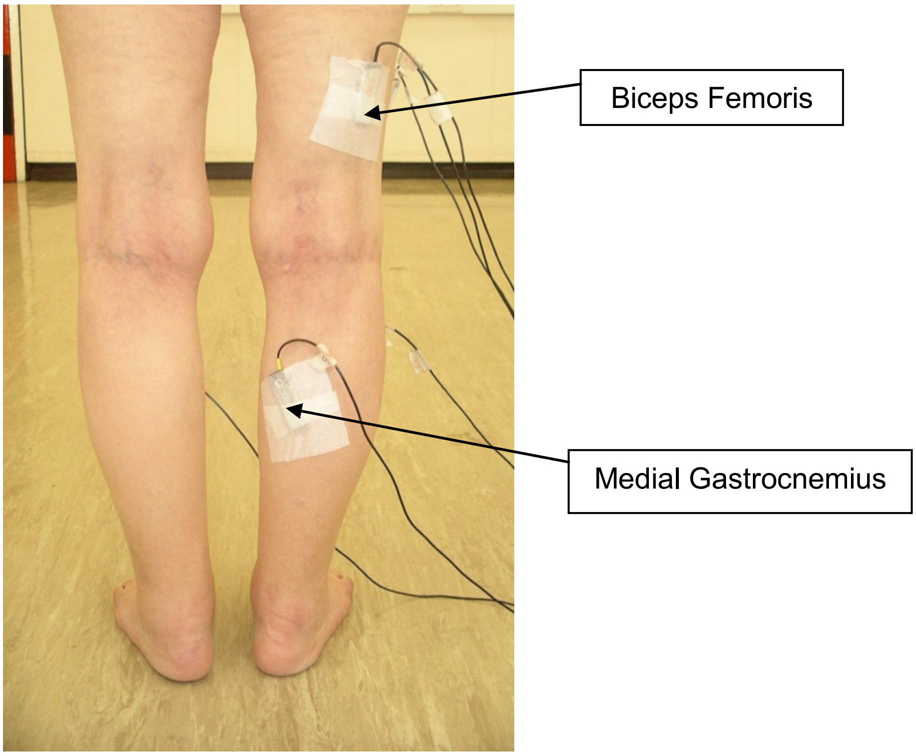

Fig 7.2: EMG electrode placement for healthy older adults – posterior view

7.6: Lower Limbs

Participants were instructed to stand in their normal standing position. Confounding factors including musculoskeletal conditions, which may include osteoarthritis, reduced range of movement and the prevalence of joint replacements at the knee in older populations, may not have permitted other methods of standardisation such as a particular knee angle.

7.7: Screening Assessments

7.7.1: Peripheral neuropathy

Older adults were screened for peripheral neuropathy using monofilaments, whilst positioned sitting upright on the plinth with their back supported and legs overlying the plinth.

7.7.2: Mini mental state examination

The MMSE was conducted on all healthy older adults to enable psychometric screening and assessment of cognitive ability prior to recruitment (Folstein et al., 1975). The maximum total score is 30 with scores of ≥27 indicating no cognitive impairment, 21–26 mild cognitive impairment, 11–20 moderate cognitive impairment and ≤10 severe cognitive impairment. Any participant scoring less than 27 was excluded from this study.

7.8: Procedures

Minor amendments to the test protocol for healthy older people included: a shorter test duration (approximately 60-90 minutes) as only one trial of quiet standing balance (per condition) was conducted (see below), inclusion of a cognitive screening test, EMG data was collected from only five lower limb muscles, parallel bars were positioned on either side of older participants as a safety precaution (Fig 7.3).

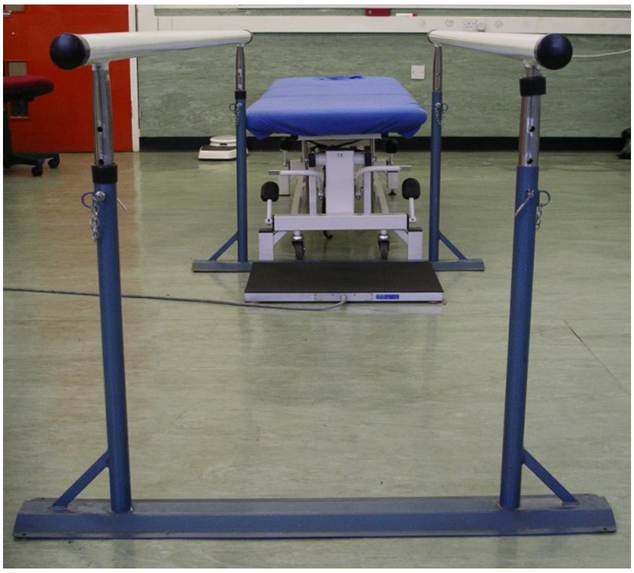

Fig 7.3: Parallel bars for healthy older adults

7.8.1: Quiet standing balance

Due to the potential for onset of fatigue, healthy older adults conducted only one repetition of quiet standing (Du Pasquier et al., 2003, Samson and Crowe, 1996) per textured surface with eyes open and one repetition with eyes closed; a total of 6 repetitions. It was considered unethical to expect older participants to conduct more extensive test procedures (Samson and Crowe, 1996).

7.9: Data Processing and Extraction

In healthy older adults, for both postural sway and EMG parameters, data from one repetition of quiet standing was used in the analysis. Upon identifying a corrupt data file, which could not be analysed, postural sway and EMG scores were calculated using data from the remaining participants.

CHAPTER 8: RESULTS FOR THE EFFECT OF TEXTURED SURFACES ON POSTURAL SWAY AND LOWER LIMB MUSCLE ACTIVITY DURING QUIET STANDING IN HEALTHY OLDER ADULTS

8.1: Introduction

This chapter will present the results for the effect of textured surfaces on postural sway parameters and lower limb EMG activity during quiet standing balance with eyes open and eyes closed, in healthy older adults. Measures of postural sway will be reported for the first 10, latter 20, and overall 30 seconds of quiet standing. Data collected simultaneously for lower limb EMG activity will be presented for quiet standing over 30 seconds under both visual conditions. Only when textured surfaces show to have a statistically significant effect on postural sway on the first 10, or latter 20 seconds of quiet standing, will corresponding EMG data be reported.

Following the methods of statistical analysis used for healthy young adults in this thesis, the distribution of data collected in healthy older adults was not explored using tests of normality. Descriptive statistics, including mean, SD, median and inter-quartile range provide insight to the distribution of data in healthy older adults and can be seen in Appendices 18-27.

8.2: Healthy Older Adults

A convenience sample of 53 healthy older adults, were recruited from the local community. Three participants from the 53 that were recruited were excluded from the study as seen in the CONSORT flow diagram (Figure 8.1). One participant reported a previous transient ischaemic attack. The second person reported labyrinthitis. The third person scored 26 on the MMSE. Table 8.1 shows descriptive characteristics of healthy older participants. Table 8.2 shows participants' self-reported medical conditions and the results of the FPI6, MMSE and monofilament screening tests.

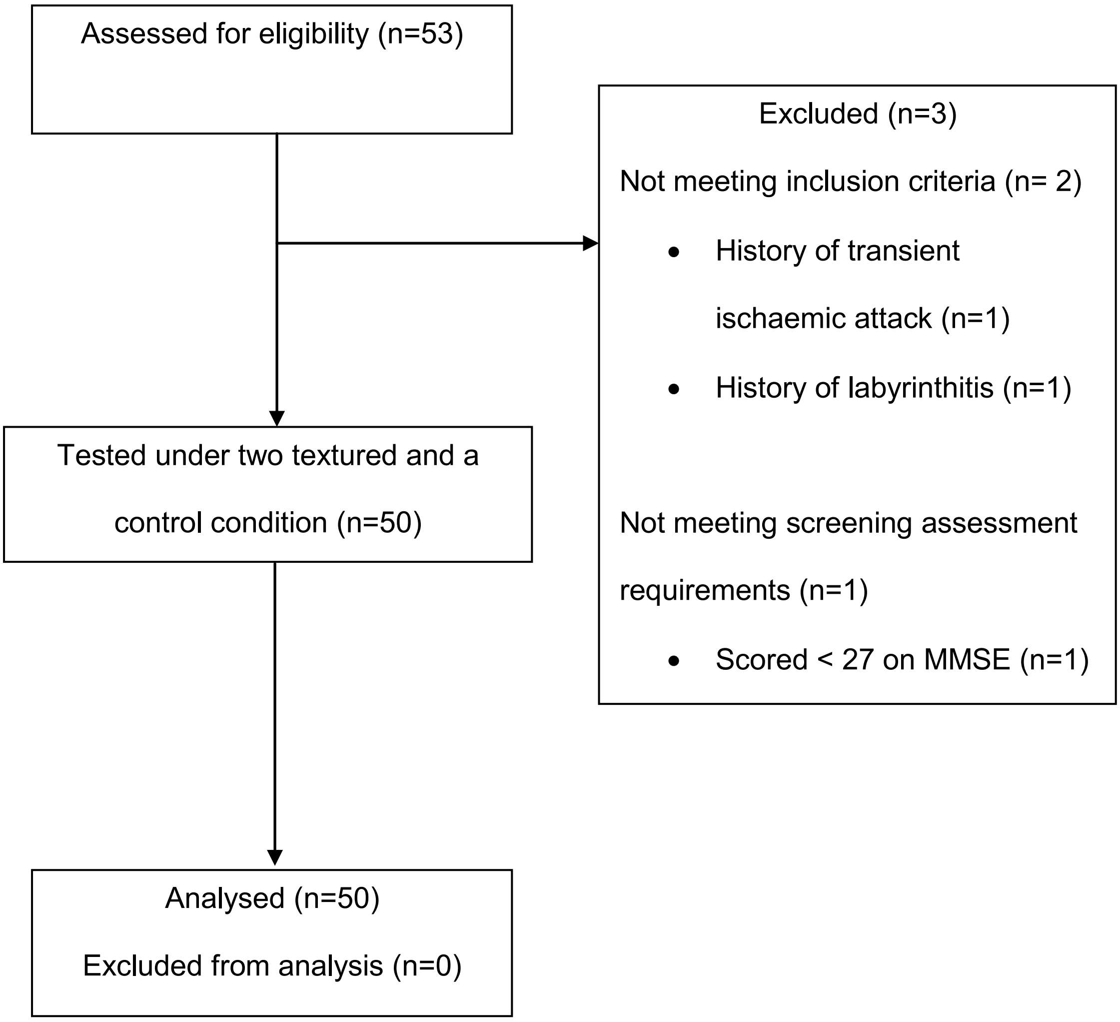

Fig 8.1: CONSORT flow diagram for healthy older adult recruitment and exclusions

Table 8.1: Descriptive characteristics of healthy older adults (n=50)

Gender	Age (yrs) Mean (SD)	Height (cm) Mean (SD)	Weight (kg) Mean (SD)	BMI (kg/m^2) Mean (SD)
21M, 29F	75.1 (5.0)	164 (7.9)	72.2 (10.9)	26.9 (4.0)

Table 8.2: Screening test results for healthy older adults (n=50)

Self-Reported Medical Conditions (n=)	FPI6 (n=)	Monofilament Testing Mean (SD) (maximum score: 10)	MMSE Score Mean (SD) (maximum score: 30)
Cardiac Bypass (4)	Normal Foot Position (42)	9.9 (0.4)	28.7 (0.9)
Other Cardiac Problems (2)	Unilateral Pronation (3)		
Diabetes (5)	Bilateral Pronation (3)		
Osteoarthritis (5)	Unilateral Supination (2)		
Hip/Knee Joint Replacements (3)			
Brain Tumour (1)			
Bowel Cancer (1)			
Sciatica (1)			
Multiple Medications (1)			

8.3: Postural Sway during Quiet Standing in Healthy Older Adults

8.3.1: Postural sway over 30 seconds with eyes open in healthy older adults

Table 8.3 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during quiet standing over 30 seconds with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CI for T1-T2 for AP SD is less conclusive. This CI for T1-T2 shows a marked asymmetry (zero close to the lower limit). This asymmetry occurs between the two textures, rather than baseline versus texture. However, there is a direct comparison in the 95% CI for C-T2 for AP SD which shows a marked asymmetry (zero close to the lower limit), in the direction of an increase in sway.

Whilst these asymmetries do not reach significance, these observations tentatively suggest the two textures may have opposite effects on AP sway. Relative to C, T2 increased AP SD by 7.4%. T1 decreased AP SD by 0.7%.

Table 8.3: Postural sway for each textured condition during quiet standing with eyes open over 30 seconds in healthy older adults (n=50)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p*=	C-T1 Mean (95%CI) *p*=	C-T1 % Change	C-T2 Mean (95% CI) *p*=	C-T2 % Change	T1-T2 Mean (95% CI) *p*=
AP SD (mm)	5.4 (1.4)	5.3 (1.5)	5.7 (1.6)	0.151	-0.04 (-0.5 to 0.5) *0.874*	-0.7	0.4 (-0.1 to 0.9) *0.125*	7.4	0.4 (-0.01 to 0.8) *0.053*
AP range (mm)	33.7 (10.5)	35.9 (10.2)	35.3 (9.4)	0.359 gg	2.2 (-1.5 to 5.9) *0.236*	6.5	1.7 (-1.8 to 5.2) *0.325*	5.0	-0.5 (-3.0 to 2.1) *0.713*
ML SD (mm)	3.4 (1.1)	3.4 (1.2)	3.3 (1.0)	0.871	0.03 (-0.2 to 0.3) *0.807*	0.9	-0.04 (-0.3 to 0.2) *0.751*	-1.2	-0.1 (-0.4 to 0.2) *0.641*
ML range (mm)	23.9 (8.0)	24.7 (8.0)	24.2 (9.0)	0.759	0.9 (-1.2 to 2.9) *0.407*	3.8	0.3 (-1.9 to 2.6) *0.781*	1.3	-0.5 (-3.2 to 2.1) *0.686*
CoP velocity (mm s^{-1})	13.5 (4.2)	13.3 (3.6)	14.8 (7.6)	0.149 gg	-0.3 (-1.2 to 0.7) *0.585*	-2.2	1.3 (-0.6 to 3.2) *0.182*	9.6	1.5 (-0.3 to 3.4) *0.104*

gg = Greenhouse-Geisser correction for sphericity

8.3.2: Postural sway over 30 seconds with eyes closed in healthy older adults

Table 8.4 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during quiet standing over 30 seconds with eyes closed.

The repeated measures ANOVAs indicated there was a statistically significant difference between the three conditions for CoP velocity (F [2,47] =5.530, *p*=0.005). Post-hoc analysis showed this significant difference occurred between T1-T2, with a mean difference (95% CI) of 2.8 mm s^{-1} (1.0 to 4.6). These observations suggest the two different textured surfaces have opposite effects on sway velocity in older adults.

An inspection of the 95% CIs for C-T1 and C-T2 support this. The CI for C-T1 for CoP velocity showed a marked asymmetry (zero close to the upper limit), in the direction of a decrease. The CI for C-T2 showed a marked asymmetry (zero close to the lower limit), in the direction of an increase in CoP velocity. Relative to C, CoP velocity decreased by 4.7% when standing on T1, and increased by 9.8% when standing on T2.

For the other sway variables, AP and ML SD, AP and ML range, the repeated measures ANOVAs indicated there were no statistically significant differences between the three conditions. However, the 95% CIs for C-T1 for AP range and ML SD are less conclusive. The CI for C-T1 for AP range does not contain zero. This uni-directional CI is in the direction of a decrease in AP sway range. Similarly, the CI for C-T1 for ML SD does not contain zero and is in the direction of a decrease in ML SD.

Whilst these uni-directional CIs did not reach significance, these observations tentatively suggest T1 may have the capacity to reduce AP and ML sway parameters. Relative to C, T1 reduced AP range by 10.1% and ML SD by 10.3%.

Table 8.4: Postural sway for each textured condition during quiet standing with eyes closed over 30 seconds in healthy older adults (n=50)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p*=	C-T1 Mean (95%CI) *p*=	C-T1 % Change	C-T2 Mean (95% CI) *p*=	C-T2 % Change	T1-T2 Mean (95% CI) *p*=
AP SD (mm)	6.1 (1.9)	5.6 (1.6)	5.8 (1.8)	0.155	-0.5 (-0.9 to 0.04) *0.073*	-8.2	-0.2 (-0.7 to 0.2) *0.298*	-3.3	0.2 (-0.2 to 0.7) *0.338*
AP range (mm)	40.4 (13.1)	36.3 (11.1)	39.2 (12.4)	0.063	-4.1 (-8.2 to -0.1) *0.047*	-10.1	-1.2 (-4.3 to 1.8) *0.417*	-3.0	2.9 (-0.6 to 6.4) *0.104*
ML SD (mm)	3.9 (1.5)	3.6 (1.5)	3.8 (1.5)	0.102	-0.4 (-0.7 to -0.03) *0.035*	-10.3	-0.1 (-0.4 to 0.2) *0.571*	-2.6	0.3 (-0.1 to 0.7) *0.162*
ML range (mm)	27.5 (9.8)	25.6 (8.8)	27.0 (9.7)	0.387	-1.9 (-4.8 to 0.9) *0.180*	-6.9	-0.6 (-3.4 to 2.3) *0.697*	-2.2	1.4 (-1.5 to 4.3) *0.351*
CoP velocity (mm s^{-1})	19.3 (7.3)	18.4 (6.4)	21.2 (8.9)	**0.005***	-0.9 (-2.3 to 0.5) *0.194*	-4.7	1.9 (-0.02 to 3.8) *0.052*	9.8	2.8 (1.0 to 4.6) *0.003*

* *p*<0.05, gg = Greenhouse-Geisser correction for sphericity

8.3.3: Postural sway over the first 10 seconds with eyes open in healthy older adults

Table 8.5 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during the first 10 seconds of quiet standing with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant. Inspection of the 95% CIs for C-T1 and C-T2 supports that. All contain zero well within the interval.

Table 8.5: Postural sway for each textured condition during quiet standing with eyes open over 10 seconds in healthy older adults (n=50)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p*=	C-T1 Mean (95%CI) *p*=	C-T1 % Change	C-T2 Mean (95% CI) *p*=	C-T2 % Change	T1-T2 Mean (95% CI) *p*=
AP SD (mm)	5.0 (1.6)	5.5 (1.7)	5.4 (1.7)	0.191	0.5 (-0.1 to 1.1) *0.119*	10.0	0.4 (-0.2 to 0.9) *0.179*	8.0	-0.1 (-0.6 to 0.4) *0.728*
AP range (mm)	28.2 (10.0)	31.4 (10.6)	30.2 (9.1)	0.108 gg	3.3 (-0.1 to 6.6) *0.058*	11.7	2.0 (-1.3 to 5.3) *0.222*	7.1	-1.2 (-3.6 to 1.2) *0.304*
ML SD (mm)	3.5 (1.3)	3.6 (1.4)	3.6 (1.3)	0.632 gg	0.1 (-0.1 to 0.4) *0.281*	2.9	0.1 (-0.2 to 0.5) *0.518*	2.9	-0.04 (-0.4 to 0.4) *0.855*
ML range (mm)	21.8 (7.5)	22.9 (8.4)	22.6 (8.8)	0.615	1.1 (-0.9 to 3.1) *0.272*	5.0	0.8 (-1.6 to 3.2) *0.485*	3.7	-0.3 (-2.9 to 2.3) *0.837*
CoP velocity (mm s^{-1})	17.4 (5.9)	17.8 (5.3)	19.2 (8.9)	0.145 gg	0.4 (-1.1 to 1.8) *0.625*	2.3	1.8 (-0.3 to 3.9) *0.092*	10.3	1.5 (-0.6 to 3.6) *0.169*

gg = Greenhouse-Geisser correction for sphericity

8.3.4: Postural sway over the first 10 seconds with eyes closed in healthy older adults

Table 8.6 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during the first 10 seconds of quiet standing with eyes closed.

The repeated measures ANOVAs indicated there was a statistically significant difference between the three conditions for CoP velocity (F [2,47] =8.611, p=<0.001). Post-hoc analysis identified these significant differences occurred between C-T2 and T1-T2. The 95% CIs for C-T2 (1.0 to 4.6) and T1-T2 (2.2 to 6.4) for CoP velocity are uni-directional and do not contain zero. The CI for C-T1 for CoP velocity also showed a marked asymmetry (zero close to the upper limit) in the direction of decreased sway velocity. These observations indicate that T2 can significantly increase CoP velocity beyond baseline, and that the two different textured surfaces have opposite effects on sway velocity. Relative to C, T1 reduced CoP velocity by 7.3%. T2 increased CoP velocity by 9.3%.

For the other sway variables, AP and ML SD, AP and ML range, the repeated measures ANOVAs indicated there were no statistically significant differences between the three conditions. However, the 95% CIs for C-T1 and T1-T2 for AP SD are less conclusive. The CI for C-T1 showed a marked asymmetry (zero close to the upper limit), in the direction of a decrease in AP SD. The CI for T1-T2 for AP SD also showed a marked asymmetry (zero close to the lower limit), in the direction of an increase. Whilst these asymmetries did not reach significance, these observations tentatively suggest both T1 and T2 can reduce AP SD, but this effect is more marked when standing on T1. Relative to C, T1 reduced AP SD by 10.9% whilst T2 reduced AP SD by 1.6%.

The 95% CIs for C-T1 and T1-T2 for AP range are also inconclusive. The CI for C-T1 showed a marked asymmetry (zero close to the upper limit), in the direction of a decrease in AP range. The CI for T1-T2 for AP range was uni-directional, in the direction of an increase. Whilst these observations did not reach significance, they tentatively suggest both T1 and T2 can reduce AP range, but this effect is more marked when standing on T1. Relative to C, T1 reduced AP range by 10.9% whilst T2 reduced AP range by 0.6%.

Table 8.6: Postural sway for each textured condition during quiet standing with eyes closed over the first 10 seconds in healthy older adults (n=50)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p=*	C-T1 Mean diff. (95%CI) *p=*	C-T1 % Change	C-T2 Mean diff. (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
AP SD (mm)	6.4 (2.7)	5.7 (1.8)	6.3 (1.9)	0.084 gg	-0.7 (-1.5 to 0.1) *0.073*	-10.9	-0.1 (-0.7 to 0.4) *0.655*	-1.6	0.6 (-0.01 to 1.1) *0.054*
AP range (mm)	35.8 (14.1)	31.9 (11.0)	35.6 (12.3)	0.066 gg	-3.9 (-8.1 to 0.4) *0.073*	-10.9	-0.2 (-3.2 to 2.8) *0.893*	-0.6	3.7 (0.2 to 7.1) *0.039*
ML SD (mm)	3.9 (1.2)	3.7 (1.3)	4.0 (1.6)	0.274	-0.2 (-0.6 to 0.1) *0.236*	-5.1	0.1 (-0.3 to 0.5) *0.643*	2.6	0.3 (-0.1 to 0.7) *0.132*
ML range (mm)	24.4 (9.5)	23.3 (8.8)	24.7 (9.3)	0.610	-1.1 (-4.0 to 1.8) *0.444*	-4.5	0.3 (-2.8 to 3.4) *0.845*	1.2	1.4 (-1.5 to 4.3) *0.340*
CoP velocity (mm s^{-1})	25.9 (10.7)	24.0 (7.9)	28.3 (10.6)	**<0.001***	-1.9 (-4.1 to 0.3) *0.091*	-7.3	2.4 (0.4 to 4.4) *0.019*	9.3	4.3 (2.2 to 6.4) *<0.001*

* $p<0.05$, gg = Greenhouse-Geisser correction for sphericity

8.3.5: Postural sway over the latter 20 seconds with eyes open in healthy older adults

Table 8.7 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during the first 20 seconds of quiet standing with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant. Inspection of the 95% CIs for C-T1 and C-T2 supports that. All contain zero well within the interval.

Table 8.7: Postural sway for each textured condition during quiet standing with eyes open over the latter 20 seconds in healthy older adults (n=50)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p=*	C-T1 Mean diff. (95%CI) *p=*	C-T1 % Change	C-T2 Mean diff. (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
AP SD (mm)	4.2 (1.4)	4.0 (1.2)	4.2 (1.4)	0.468	-0.2 (-0.7 to 0.2) *0.330*	-4.8	0.03 (-0.4 to 0.5) *0.911*	0.7	0.2 (-0.2 to 0.7) *0.254*
AP range (mm)	22.9 (6.6)	22.3 (6.1)	22.7 (6.1)	0.806	-0.6 (-2.8 to 1.5) *0.558*	-2.6	-0.2 (-2.0 to 1.7) *0.850*	-0.9	0.5 (-1.5 to 2.5) *0.643*
ML SD (mm)	2.7 (0.9)	2.6 (1.0)	2.6 (0.8)	0.712 gg	-0.1 (-0.2 to 0.1) *0.332*	-3.7	-0.05 (-0.3 to 0.2) *0.674*	-1.9	0.03 (-0.2 to 0.3) *0.782*
ML range (mm)	16.7 (5.4)	15.9 (4.3)	15.8 (4.1)	0.284	-0.8 (-2.1 to 0.5) *0.203*	-4.8	-0.9 (-2.3 to 0.5) *0.197*	-5.4	-0.1 (-1.2 to 1.0) *0.869*
CoP velocity (mm s^{-1})	11.6 (3.7)	11.0 (3.2)	12.6 (7.3)	0.173 gg	-0.5 (-1.4 to 0.3) *0.181*	-4.3	1.0 (-0.9 to 3.0) *0.293*	8.6	1.6 (-0.4 to 3.5) *0.110*

gg = Greenhouse-Geisser correction for sphericity

8.3.6: Postural sway over the latter 20 seconds with eyes closed in healthy older adults

Table 8.8 shows the pairwise comparisons, magnitude and direction of change in postural sway parameters between conditions during the latter 20 seconds of quiet standing with eyes closed.

The repeated measures ANOVAs indicated there was a statistically significant difference between the three conditions for ML range (F [2,47] =3.840, p=0.033). Post-hoc analysis identified this significant difference occurred between C-T1. The 95% CI for C-T1 for ML range is uni-directional, in the direction of a decrease in ML range. Relative to C, T1 reduced ML range by 9.2%.

For the other sway variables, AP and ML SD, AP range and CoP velocity, the repeated measures ANOVAs indicated there were no statistically significant differences between the three conditions. However, the 95% CI for C-T1 for ML SD is less conclusive. The CI for C-T1 for ML SD is also uni-directional, in the direction of a decrease in ML SD. Whilst this observation does not reach significance it tentatively suggests T1 can reduce ML SD beyond baseline. Relative to C, T1 reduced ML SD by 10%.

The 95% CI for C-T2 for CoP velocity is less conclusive. The CI for C-T2 for CoP velocity showed a marked asymmetry (zero close to the lower limit), in the direction of an increase. Relative to C, T2 increased CoP velocity by 10%. The CI for T1-T2 for CoP velocity does not contain zero. Whilst these asymmetries did not reach significance, these observations tentatively suggest T2 may have the capacity to increase CoP velocity and that the two different textures may have opposite effects on sway velocity.

Table 8.8: Postural sway for each textured condition during quiet standing with eyes closed over the latter 20 seconds in healthy older adults (n=50)

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	ANOVA *p=*	C-T1 Mean diff. (95%CI) *p=*	C-T1 % Change	C-T2 Mean diff. (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
AP SD (mm)	4.9 (1.7)	4.6 (1.6)	4.6 (1.5)	0.371	-0.3 (-0.8 to 0.2) *0.251*	-6.1	-0.2 (-0.6 to 0.1) *0.215*	-4.1	0.05 (-0.4 to 0.5) *0.834*
AP range (mm)	27.3 (9.1)	25.8 (7.6)	26.4 (8.8)	0.518	-1.5 (-4.1 to 1.2) *0.277*	-5.5	-0.8 (-3.4 to 1.8) *0.534*	-2.9	0.6 (-1.7 to 3.0) *0.588*
ML SD (mm)	3.0 (1.3)	2.7 (1.0)	2.9 (1.1)	0.077	-0.3 (-0.5 to -0.02) *0.036*	-10.0	-0.1 (-0.3 to 0.1) *0.209*	-3.3	0.1 (-0.1 to 0.3) *0.278*
ML range (mm)	18.5 (7.3)	16.7 (4.7)	17.4 (5.5)	**0.033 gg***	-1.7 (-3.3 to -0.2) *0.027*	-9.2	-1.1 (-2.3 to 0.1) *0.067*	-5.9	0.6 (-0.4 to 1.7) *0.248*
CoP velocity (mm s^{-1})	16.0 (6.7)	15.6 (6.1)	17.6 (8.7)	0.096 gg	-0.4 (-1.9 to 1.1) *0.565*	-2.5	1.6 (-0.6 to 3.8) *0.149*	10.0	2.0 (0.03 to 4.0) *0.047*

* p<0.05, gg = Greenhouse-Geisser correction for sphericity

8.4: EMG during Quiet Standing in Healthy Older Adults

8.4.1: EMG activity over 30 seconds with eyes open in healthy older adults

Table 8.9 shows the pairwise comparisons, magnitude and direction of change in lower limb EMG activity between conditions during quiet standing over 30 seconds with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant. Inspection of the 95% CIs for C-T1 and C-T2 supports that. All contain zero well within the interval.

It is worth noting that the SD for EMG activity in VL and BF considerably exceed the mean amplitude, for all three surfaces (Table 8.9). This is also seen in MG when standing on T1. Mean EMG amplitude for VL, BF and MG was calculated using samples of less than 50 participants: 46 (VL), 46 (BF) and 49 (MG). This was due to corrupt EMG data which had to be excluded from the analysis.

Table 8.9: Lower limb EMG activity (μV) for each textured condition during quiet standing with eyes open over 30 seconds in healthy older adults

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	n=	ANOVA *p=*	C-T1 Mean (95%CI) *p=*	C-T1 % Change	C-T2 Mean (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
Rectus Femoris	12.4 (4.1)	12.3 (4.1)	12.6 (4.9)	50	0.678	-0.1 (-0.7 to 0.5) *0.726*	-0.8	0.2 (-0.5 to 0.9) *0.578*	1.6	0.3 (-0.5 to 1.1) *0.435*
Vastus Lateralis	14.5 (15.6)	15.1 (16.9)	14.8 (17.0)	46	0.845 gg	0.5 (-1.1 to 2.1) *0.511*	3.4	0.3 (-2.0 to 2.5) *0.823*	2.1	-0.3 (-2.9 to 2.4) *0.845*
Biceps Femoris	6.0 (10.4)	6.4 (9.9)	8.3 (15.5)	46	0.190 gg	0.5 (-1.3 to 2.2) *0.610*	8.3	2.3 (-0.8 to 5.3) *0.140*	38.3	1.8 (-1.0 to 4.7) *0.198*
Tibialis Anterior	3.4 (3.4)	3.1 (2.9)	3.6 (3.7)	50	0.187	-0.4 (-0.8 to 0.1) *0.150*	-11.8	0.1 (-0.5 to 0.7) *0.665*	2.9	0.5 (-0.1 to 1.0) *0.081*
Medial Gastrocnemius	6.9 (5.7)	7.9 (8.8)	7.6 (7.0)	49	0.163 gg	1.1 (-0.4 to 2.5) *0.138*	15.9	0.8 (-0.2 to 1.7) *0.108*	11.6	-0.3 (-1.3 to 0.7) *0.529*

gg = Greenhouse-Geisser correction for sphericity

8.4.2: EMG activity over 30 seconds with eyes closed in healthy older adults

Table 8.10 shows the pairwise comparisons, magnitude and direction of change in lower limb EMG activity between conditions during quiet standing over 30 seconds with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CI for C-T1 for TA activity is less conclusive. This CI for C-T1 showed a marked asymmetry (zero close to the upper limit) in the direction of a decrease in TA activity. Whilst this asymmetry did not reach significance, this observation suggests T1 may have the capacity to reduce TA activity. Relative to C, T1 reduced TA activity by 18.8%. Reductions in the amplitude of TA activity when standing on T1 may contribute to changes observed in postural sway variables during these same test conditions (Table 8.4).

Table 8.10: Lower limb EMG activity (µV) for each textured condition during quiet standing with eyes closed over 30 seconds in healthy older adults

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	n=	ANOVA *p=*	C-T1 Mean diff. (95%CI) *p=*	C-T1 % Change	C-T2 Mean diff. (95% CI) *p=*	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p=*
Rectus Femoris	13.5 (4.9)	13.0 (5.0)	13.3 (4.8)	50	0.453	-0.6 (-1.5 to 0.3) *0.184*	-4.4	-0.3 (-1.1 to 0.5) *0.485*	-2.2	0.3 (-0.8 to 1.4) *0.583*
Vastus Lateralis	16.5 (15.2)	16.0 (17.8)	17.2 (17.7)	46	0.667 gg	-0.5 (-3.2 to 2.3) *0.736*	-3.0	0.7 (-1.5 to 2.8) *0.534*	4.2	1.1 (-2.5 to 4.8) *0.537*
Biceps Femoris	6.4 (9.9)	6.2 (9.3)	6.0 (9.6)	48	0.893	-0.2 (-2.1 to 1.8) *0.854*	-3.1	-0.4 (-2.0 to 1.1) *0.573*	-6.3	-0.3 (-2.4 to 1.8) *0.802*
Tibialis Anterior	6.4 (8.6)	5.2 (7.3)	6.0 (7.7)	50	0.260	-1.2 (-2.8 to 0.4) *0.139*	-18.8	-0.4 (-1.7 to 0.9) *0.513*	-6.3	0.8 (-0.8 to 2.3) *0.313*
Medial Gastrocnemius	8.9 (7.4)	8.8 (8.2)	9.3 (7.5)	49	0.614	-0.1 (-1.1 to 0.9) *0.833*	-1.1	0.4 (-0.7 to 1.5) *0.489*	4.5	0.5 (-0.6 to 1.5) *0.351*

gg = Greenhouse-Geisser correction for sphericity

8.4.3: EMG activity over the first 10 seconds with eyes closed in healthy older adults

Table 8.11 shows the pairwise comparisons, magnitude and direction of change in lower limb EMG activity between conditions during quiet standing over 30 seconds with eyes open. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CI for C-T1 for TA activity is less conclusive. This CI for C-T1 showed a marked asymmetry (zero close to the upper limit) in the direction of a decrease in TA activity. Whilst this asymmetry did not reach significance, this observation suggests T1 may have the capacity to reduce TA activity. Relative to C, T1 reduced TA activity by 16.4%. Reductions in the amplitude of TA activity when standing on T1 may contribute to changes observed in postural sway variables during these same test conditions (Table 8.6).

Table 8.11: Lower limb EMG activity (μV) for each textured condition during quiet standing with eyes closed over the first 10 seconds in healthy older adults

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	n=	ANOVA *p*=	C-T1 Mean diff. (95%CI) *p*=	C-T1 % Change	C-T2 Mean (95% CI) *p*=	C-T2 % Change	T1-T2 Mean (95% CI) *p*=
Rectus Femoris	29.0 (5.3)	28.6 (5.9)	28.7 (5.2)	50	0.689 gg	-0.4 (-1.3 to 0.6) *0.417*	-1.4	-0.3 (-1.1 to 0.6) *0.519*	-1.0	0.1 (-1.1 to 1.3) *0.833*
Vastus Lateralis	17.5 (16.7)	17.9 (21.5)	18.2 (18.6)	46	0.832 gg	0.4 (-2.7 to 3.5) *0.796*	2.3	0.8 (-1.6 to 3.2) *0.527*	4.6	0.4 (-3.8 to 4.6) *0.864*
Biceps Femoris	6.5 (10.2)	6.4 (10.6)	6.8 (11.2)	48	0.935	-0.1 (-2.3 to 2.0) *0.902*	-1.5	0.2 (-1.4 to 1.9) *0.782*	3.1	0.4 (-1.8 to 2.5) *0.738*
Tibialis Anterior	6.7 (9.7)	5.6 (8.1)	6.1 (8.9)	50	0.308 gg	-1.1 (-2.7 to 0.4) *0.145*	-16.4	-0.6 (-2.3 to 1.1) *0.494*	-9.0	0.6 (-0.6 to 1.8) *0.351*
Medial Gastrocnemius	8.8 (7.7)	8.3 (7.2)	8.8 (6.9)	49	0.528	-0.5 (-1.5 to 0.5) *0.308*	-5.7	-0.02 (-1.1 to 1.1) *0.976*	-0.2	0.5 (-0.4 to 1.4) *0.281*

gg = Greenhouse-Geisser correction for sphericity

8.4.4: EMG activity over the latter 20 seconds with eyes closed in healthy older adults

Table 8.12 shows the pairwise comparisons, magnitude and direction of change in lower limb EMG activity between conditions over the latter 20 seconds of quiet standing with eyes closed. The repeated measures ANOVAs showed these differences were not statistically significant.

However, the 95% CI for C-T1 for TA activity is less conclusive. This CI for C-T1 showed a marked asymmetry (zero close to the upper limit) in the direction of a decrease in TA activity. Whilst this asymmetry did not reach significance, this observation suggests T1 may have the capacity to reduce TA activity. Relative to C, T1 reduced TA activity by 19%. Reductions in the amplitude of TA activity when standing on T1 may contribute to changes observed in postural sway variables during these same test conditions (Table 8.8).

Table 8.12: Lower limb EMG activity (μV) for each textured condition during quiet standing with eyes closed over the latter 20 seconds in healthy older adults

	C Mean (SD)	T 1 Mean (SD)	T 2 Mean (SD)	n=	ANOVA *p*=	C-T1 Mean diff. (95%CI) *p*=	C-T1 % Change	C-T2 Mean diff. (95% CI) *p*=	C-T2 % Change	T1-T2 Mean diff. (95% CI) *p*=
Rectus Femoris	5.8 (4.8)	5.1 (4.6)	5.5 (4.6)	50	0.341	-0.7 (-1.5 to 0.2) *0.119*	-12.1	-0.3 (-1.2 to 0.6) *0.479*	-5.2	0.4 (-0.7 to 1.4) *0.476*
Vastus Lateralis	16.0 (14.8)	15.1 (16.3)	16.7 (17.4)	46	0.505 gg	-0.9 (-3.6 to 1.8) *0.503*	-5.6	0.6 (-1.5 to 2.7) *0.549*	3.8	1.5 (-2.0 to 5.0) *0.384*
Biceps Femoris	6.4 (10.0)	6.2 (8.9)	5.6 (8.9)	48	0.690	-0.2 (-2.1 to 1.7) *0.831*	-3.1	-0.8 (-2.4 to 0.8) *0.325*	-12.5	-0.6 (-2.7 to 1.6) *0.591*
Tibialis Anterior	6.3 (8.8)	5.1 (7.4)	5.5 (7.2)	50	0.340	-1.2 (-3.1 to 0.6) *0.184*	-19.0	-0.9 (-2.4 to 0.7) *0.272*	-14.3	0.4 (-1.4 to 2.1) *0.667*
Medial Gastrocnemius	9.0 (7.5)	9.1 (8.9)	9.6 (7.9)	49	0.570	0.1 (-1.1 to 1.2) *0.897*	1.1	0.6 (-0.6 to 1.8) *0.340*	6.7	0.5 (-0.7 to 1.7) *0.406*

gg = Greenhouse-Geisser correction for sphericity

8.5: Summary

The experimental effect of textured surfaces on postural sway and lower limb muscle activity in healthy older adults was explored during tests of quiet standing balance with eyes open and eyes closed.

Findings from the current study suggest textured surfaces may have a beneficial effect on balance performance in healthy older adults. Based on the general agreement that a reduction in the distance or velocity of CoP movement represents improved postural stability, findings from the current study suggest the textured pattern specific to T1 brought about such an effect.

Relative to C, T1 showed to significantly reduce ML sway range and CoP velocity when healthy older adults stood quietly with their eyes closed. Inspection of the 95% CIs for pairwise comparisons tentatively suggests T1 can also reduce AP SD, ML SD and AP range: these observations did not reach significance.

Relative to C, T2 showed to significantly increase CoP velocity during quiet standing with eyes closed over 10 seconds in older adults. This may suggest the textured pattern specific to T2, could have a detrimental effect on balance performance in healthy older adults.

Significant effects of texture on postural sway parameters were only observed when older adults stood quietly with their eyes closed. Textured surfaces had no effect on postural sway during eyes open trials. This suggests enhanced sensory feedback provided by a textured surface, may be used as a surrogate source of sensory information for balance control, in the absence of visual inputs. This concept will be discussed in chapter 9 of the current thesis.

This study set out to determine whether textured surfaces could change postural sway in healthy older adults and whether such changes were accompanied by alterations in lower limb muscle activity, as a potential underlying mechanism. The findings from healthy older adults indicate that alterations in sway parameters may be accompanied by small changes in lower leg muscle activity, specifically TA.

CHAPTER 9: DISCUSSION OF THE EFFECT OF TEXTURED SURFACES ON POSTURAL SWAY AND LOWER LIMB MUSCLE ACTIVITY DURING QUIET STANDING IN HEALTHY OLDER ADULTS

9.1: Introduction

This chapter will discuss the findings from healthy older adults within the context of previous literature surrounding postural control, muscular contributions to upright standing, and the use of footwear interventions for improving balance in healthy older people.

9.2: Textured Surfaces and Postural Sway in Healthy Older Adults

The current study showed that textured surfaces can significantly change the magnitude and velocity of CoP movement in healthy older adults during quiet standing. Standing on a textured surface with eyes closed, had a statistically significant effect on ML range and CoP velocity. No significant differences in postural sway were reported between the textured conditions, when standing with eyes open. Neither was there any effect on AP sway in both visual conditions. In the current study, improvements in balance were recognised as a reduction in the distance of CoP movement in either AP or ML directions, or CoP velocity, compared to baseline (Judge et al., 1993, Nagy et al., 2007).

Table 9.1 shows control data from the current study, compared to measures of quiet standing reported in previous research, under similar test conditions (Laughton et al., 2003, Mackey and Robinovitch, 2005, Melzer et al., 2004, Raymakers et al., 2005, Stel et al., 2003). Comparisons were only made with other studies which collected sway data from a force platform, during unperturbed quiet standing with eyes open, whilst standing on a flat, rigid surface.

Table 9.1 indicates that healthy older adults in the current study did not show a similar magnitude of AP or ML sway and CoP velocity to previous work. This suggests data for baseline quiet standing balance in healthy older adults is highly variable throughout literature, and is specific to the sample being investigated. This evidence strengthens the research design of the current thesis, indicating that older adults should be tested under both textured and no texture conditions, thus acting as their own controls, rather than in comparison to a control group.

Table 9.1: Comparative baseline postural sway data during quiet standing with eyes open on a flat, rigid surface in healthy older adults between the current and previous studies

Study	**Participants**	**Test Conditions**	**AP Sway (mm) Mean (SD)**	**ML Sway (mm) Mean (SD)**	**CoP Velocity (mm s^{-1}) Mean (SD)**
Current study	Healthy older adults	Quiet standing, 30 seconds, eyes open	*APSD :* 5.4 (1.4) *AP Range :* 33.7 (10.5)	*MLSD :* 3.4 (1.1) *ML Range :* 23.9 (8.0)	*CoP Velocity :* 13.5 (4.2)
Laughton et al. (2003)	Healthy older adults, non-fallers	Quiet standing, 30 seconds, eyes open	*APSD:* 4.2 (1.1) *AP Range:* 20.7 (1.3)	*MLSD:* 2.9 (1.3) *ML Range:* 14.2 (6.4)	-
Stel et al. (2003)	Elderly non- and once-fallers	Quiet standing, 30 seconds, eyes open	*AP sway:* 4.1 (2.7)	*ML sway:* 3.4 (1.3)	-
Melzer et al. (2004)	Elderly non-fallers	Quiet standing, 20 seconds, eyes open	*AP sway:* 25.0 (0.8)	*ML sway:* 29.0 (0.9)	*CoP Velocity:* 19.0 (1.0)
Mackey and Robinovitch (2005)	Healthy older females	Quiet standing, 15 seconds, eyes open	*AP Range:* 17.0 (6.2)	*ML Range:* 8.8 (4.9)	-
Raymakers et al. (2005)	Healthy older adults	Quiet standing, 60 seconds, eyes open	*AP Range:* 25.0 (8.0)	*ML Range:* 24.0 (7.0)	*CoP Velocity:* 15.9 (11.5)

The findings from healthy older adults indicate that T1 and T2 had statistically significant opposite effects on CoP velocity during the first 10, and overall 30 seconds, of quiet standing with eyes closed. The clinical importance of this finding can be highlighted by exploring changes in CoP velocity within the context of baseline versus texture comparisons. During the first 10 seconds of quiet standing with eyes closed, T2 significantly increased CoP velocity beyond control. Under these same test conditions, standing on T1 showed a tentative reduction in CoP velocity. This evidence suggests a textured surface comprising convex circular patterning may be detrimental to balance performance. This is based on the assumption if the CoP changes its position at a faster velocity, then the CoM may reach, and possibly exceed, the limits of stability much quicker, resulting in less time to make postural corrections. Therefore, in the absence of visual information, it appears that the material properties, geometry and spacing of protrusions exclusive to T1 may reflect an 'optimal' textured design, capable of enhancing upright standing balance.

There is very little previous research exploring the effect of texture on balance performance in older adults. Given the relationship between increasing age and deterioration in postural control, the potential for textured interventions to enhance balance in a vulnerable older population has received little attention. To date, only one study has explored the effect of textured foot insoles on postural sway in healthy older adults (Palluel et al., 2008). Therefore, the findings from the current study are novel and address an important gap in current literature.

Collectively, the results from the current study and those of Palluel et al. (2008) indicate textured surfaces and foot insoles can improve balance in older adults. Palluel et al. (2008) reported an improvement in postural stability over 5 minutes during standing with spike foot insoles. However, no significant changes in postural stability were observed upon 'immediate' exposure (the first 32 seconds) to the insoles (Palluel et al., 2008). This conflicts with the current findings and may be attributable to the duration of data acquisition in each study. The current study reported that during the initial 10 seconds of quiet standing with eyes closed, textured surfaces significantly altered CoP velocity and may also have the potential to change AP sway in healthy older adults. As discussed in chapter 2 of this thesis, sway data can differ over 10, 20 and 30 seconds (Le Clair and Riach, 1996). Therefore, it is possible that the immediate effect of spike insoles on balance control may have been overlooked within the first 32 seconds of sway data in the study by Palluel et al. (2008).

Relative to baseline measures, T2 appears to increase sway parameters in healthy older adults, thus suggesting T2 may not act as a surrogate source of sensory information when visual information is removed. Whilst this direction of change in sway parameters may be viewed as clinically undesirable, such evidence demonstrates the CNS has recognised, and is responding to, enhanced sensory feedback provided by this textured surface. It is also possible that increasing the distance or velocity of CoP movement may actually serve to increase cutaneous sensory input from the feet and proprioceptive information from stretch receptors in the lower limb muscles (Benjuya et al., 2004). This may have a de-stabilising effect in the short-term, but potentially enhance long-term balance ability. This remains unknown and requires further investigation.

It is possible that the influence of textured surfaces on balance performance may not take immediate effect in all older people: some may require a short period of acclimatisation to a new stimulus. This would indicate that individuals may respond to enhanced plantar sensory information at different rates. Alternatively, delayed detection or processing of sensory information for balance control may represent varying levels of physiological functioning within the current study sample, which spanned three age decades.

Whilst CoP velocity is considered to be a sensitive and repeatable sway variable (Lafond et al., 2004, Lin et al., 2008, Raymakers et al., 2005, Takala et al., 1997), it is unlikely this single sway parameter provides a clear picture of the interaction between texture and balance performance.

During the latter 20 seconds of quiet standing with eyes closed, a statistically significant effect was seen for ML sway range between C and T1. Textured surfaces did not alter lateral sway parameters during quiet standing over the first 10 or overall 30 seconds. Such evidence may substantiate the need to discard initial sway data to ensure reliable analysis. The first 10 seconds of sway data may be highly variable (Le Clair and Riach, 1996, Raymakers et al., 2005) and as such could mask any true textured effect and contaminate sway data collected over 30 seconds. This may suggest that in the current study textured surfaces only had a true effect on ML range. The significant effects of texture on CoP velocity, reported during the first 10 and overall 30 seconds of quiet standing tests could reflect large sway variability originating from participant or force platform stabilisation, rather than a true textured effect. However, it was vital to assess the first 10 seconds of sway data in the current thesis; discarding such data may lead to loss of important information, particularly in older adults. During the first few seconds after standing up from a seated position, older adults are at risk of losing their balance, whilst the body is re-stabilising. Therefore, it would be desirable for textured surfaces to take effect when the body is experiencing postural instability.

In the current study, an exploratory sub-group analysis for gender was carried out because previous research suggests older females may be more prone to postural instability compared to older males (Gbiri and Fabunmi, 2006). Differences in balance ability between older males and females could be attributed to a number of gender-specific physiological factors including declines in neuromuscular function, muscle strength, muscle mass, and bone density associated with the menopausal period in females (Brooke-Wavell et al., 2001, Trudelle-Jackson et al., 2006).

Speculative findings from the sub-group analysis for gender suggested that textured surfaces may influence a larger spectrum of sway parameters in healthy older females, compared to older males. For more information see Appendices 28-29.

Table 9.2 provides a summary for the effects of textured surfaces on sway parameters in healthy older females. This table reports only the sway parameters shown to be significantly affected by the textured surfaces, and identifies between which pair of textures these differences occurred. The parentheses display the direction of change in sway for each texture, relative to control. The significant effects of textured surfaces on postural sway parameters in older females were largely observed when standing with eyes closed. This supports previous evidence indicating that more challenging balance tests which substantially threaten upright stability may be more sensitive in detecting gender differences (Butler et al., 2009, Gbiri and Fabunmi, 2006, Kinney La Pier et al., 1997).

Table 9.2: Summary of where significant differences occurred between pairs of conditions for sway parameters in healthy older females (n=29). Arrows indicate the direction of change in sway when standing on Texture 1 and Texture 2, relative to Control.

	Eyes Open	Eyes Closed		
	30 seconds	*10 seconds*	*20 seconds*	*30 seconds*
AP SD	T1 vs T2 (↓ *T1*, ↑ *T2*)	T1 vs T2 (↓ *T1*, ↑ *T2*)	-	C vs T1 (↓ *T1*)
AP range	-	T1 vs T2 (↓ *T1*, ↑ *T2*)	-	C vs T1 (↓ *T1*) T1 vs T2 (↓ *T1*, ↑ *T2*)
ML range	-	-	C vs T1 (↓ *T1*)	-
CoP velocity	-	C vs T2 (↑ *T2*) T1 vs T2 (↓ *T1*, ↑ *T2*)	-	C vs T1 (↓ *T1*) T1 vs T2 (↓ *T1*, ↑ *T2*)

↓= *Texture 1/Texture 2 reduced sway relative to Control*

↑= *Texture 1/Texture 2 increased sway relative to Control*

Sub-group analysis for male participants reported only one significant finding. During the first 10 seconds of quiet standing with eyes closed, a significant difference was observed for CoP velocity between T1 and T2. However, this finding was not exclusive to male participants, but also reported as being significant in female older adults and the overall mixed-gender sample.

The current findings for gender effects also support the assumption that there may be an optimal textured pattern for enhancing standing balance. Synonymous with the results for the mixed gender sample, T1 reduced, whilst T2 increased sway parameters in females.

There is limited previous research to support the findings for the gender effects of textured surfaces in the current study. To date, only two studies (Watanabe and Okubo, 1981, Wilson et al., 2008) have explored the effect of texture on balance performance using single gender study samples. As discussed in chapter 2 of the current thesis, Watanabe and Okubo (1981) explored the effect of a textured surface on postural stability in healthy young males. Wilson et al. (2008) investigated the effect of textured insoles on balance performance in healthy middle-aged females. To date, no study has explored the gender effects of texture on older adults.

The findings from the sub-group analysis for gender generated exploratory data suggesting that textured surfaces may have greater therapeutic benefit on postural stability in older females, compared to older males. It is possible that corrective postural responses may be less efficient in older females compared to males. Postural control mechanisms, including the processing of sensory information such as tactile plantar stimulation in older adults, may be gender-specific. These concepts require further investigation.

9.3: Textured Surfaces and Lower Limb EMG Activity in Healthy Older Adults

To date, no study has investigated the effect of textured surfaces or insoles on lower limb EMG activity in older adults. Previous research investigating the effect of texture on lower limb muscle activity was specific to healthy young adults (Nurse et al., 2005) and adults with Multiple Sclerosis (Kelleher et al., 2010). Both these studies reported that textured insoles can significantly alter lower limb muscle activity during walking.

The current study reported that significant changes in postural sway parameters, resulting from older adults standing on a textured surface are not accompanied with significant changes in the amplitude of lower limb EMG activity. This suggests alterations in the intensity of lower limb muscle activity may not be underlying mechanisms to changes in balance performance. There are a number of factors which could explain the absence of significant changes in lower limb EMG activity including the; muscles of interest, method of EMG analysis, nature of the balance task and preferred mechanism of balance control in older adults.

In healthy older adults, the lower limb muscles of interest were RF, VL, BF, TA and MG. It is possible significant changes in lower limb muscle activity may have occurred in other muscles not investigated, such as the hip abductors and adductors, or ankle invertors and evertors. This is particularly relevant when considering the significant effect of textured surfaces on ML sway range: lateral CoP movement predominantly occurs through hip adductor and abductor activity (Winter, 1995).

The current study explored the effect of texture on the average amplitude of lower limb muscle activity, calculated using the average rectified value. This is a standard, robust measure of EMG activity, used in previous studies (Edwards et al., 2008) and recommended by SENIAM for the analysis of non-dynamic contractions (Merletti, 1999). Previous literature has explored the effect of footwear interventions on others components of the EMG signal including frequency wavelet analysis (Mundermann et al., 2006, Nurse et al., 2005, Rose et al., 2002, Vanicek et al., 2004), onset timing (Bird et al., 2003, Tomaro and Burdett, 1993) and co-contraction patterns (Rose et al., 2002). It is possible that calculating EMG amplitude, by averaging EMG activity over 30 seconds, may result in smoothing out any significant changes occurring in lower limb muscle activity between textured conditions. Subsequently, alternative techniques of EMG analysis may provide a broader insight relating to the effect of textured surfaces on sensorimotor function. Each measure would require further in depth investigation, which should be addressed in future research.

Tests of quiet standing may not have sufficiently threatened balance in healthy older people, to a degree whereby compensatory lower limb muscle activity was required. Therefore, given the unchallenging nature of quiet standing and the firm density of the control and textured surfaces, this would suggest any alterations in lower limb EMG activity would most likely have been observed in TA and MG. This is based on the premise that the ankle strategy is used to maintain upright balance, during unperturbed standing on a firm surface (Shumway-Cook and Woollacott, 1995).

No statistically significant changes were reported for lower limb muscle activity in healthy older adults when standing on textured surfaces. However, what can be seen is a consistent trend for reduced TA activity when standing on T1 compared to C. This change in muscle activity did not reach significance but is tentatively reflected in the distribution of scores with the 95% CI for pairwise comparisons. Reducing lower limb muscle activity may reduce noise-like fluctuations at the lower limb joints (Nagy et al., 2007). Within the context of the current study minimising muscle activity could reduce CoP movement and thus account for the short-term decreases observed in ML sway and CoP velocity when standing on T1 (Nagy et al., 2007).

The results from the current study show a similar trend to the findings from Nurse et al. (2005) which suggest textured interventions can alter EMG activity in lower leg muscles. Nurse et al. (2005) concluded that textured insoles significantly reduce TA and SOL activity in healthy young adults during walking. TA plays a major role in controlling AP CoP movement, is the strongest dorsiflexor of the foot (Moore and Dalley, 1999), but also functions as an ankle inverter, and may therefore have a minor role in ML balance control (Winter, 1995). During walking, TA activity is largely responsible for ankle dorsiflexion to permit toe clearance. Poor toe clearance, originating from TA weakness or fatigue, increases the risk of falling in older people (Menz et al., 2006). Therefore, should textured surfaces be capable of reducing TA activity, without being detrimental to static or dynamic balance performance, a textured footwear intervention could enhance TA fatigue resistance during prolonged standing or allow more frequent, successful toe clearance during locomotion. This requires further investigation.

An exploratory sub-group analysis for gender effects of texture on lower limb muscle activity was carried out. This speculative analysis showed that the two different textured surfaces had significantly opposite effects on MG activity during the latter 20 seconds of quiet standing with eyes closed, in female older adults. For more information see Appendix 30.

Kelleher et al. (2010) reported that textured insoles can significantly increase medial and lateral gastrocnemius activity in adults with Multiple Sclerosis during walking. In comparison to the current exploratory sub-group analysis for gender, this points to two suggestions. Firstly, the textured insoles used by Kelleher et al. (2010) were constructed from sandpaper. It is possible that this texture brings about a similar effect to that of T2 in the current study. Both T2 and the sandpaper appeared to increase activity in gastrocnemius. Secondly, the significant increase in gastrocnemius activity in the study by Kelleher et al. (2010) may reflect underlying abnormal neuromuscular responses to tactile stimulation, due to the presence of pathology: older adults in the current study were healthy.

There is limited research investigating gender differences in the contributions of lower limb muscle activity to postural control. Thus, it is difficult to hypothesise why textured surfaces significantly altered MG activity in female participants. The sub-group analysis for gender was purely exploratory and it is possible this finding may represent a random event. Inconsistent patterns of synergistic lower limb muscle activity can occur during balance tasks, due to variations in the amplitude of neural commands originating from the CNS (Torres-Oviedo and Ting, 2007). These authors explored lower limb muscle synergies associated with postural control during force platform perturbations (Torres-Oviedo and Ting, 2007). In healthy young adults, patterns of synergistic muscle activity showed trial-to-trial variations. Should this also be true for older adults, it is possible that the significant effect of textured surfaces on MG activity, observed during only one test condition, may have occurred by chance. This would also fit with observations that older people can use mixed ankle and hip strategies to maintain upright balance (Amiridis et al., 2003). The gender effects of textured surfaces on muscle activation patterns in older adults, requires further investigation.

In the current study, it is possible that textured surfaces did not significantly change the amplitude of activity in a greater number of muscles or test conditions, due to the low levels of muscle activity required for maintaining upright posture. Panzer et al. (1995) analysed the contributions of trunk and lower limb muscle activity, including BF, TA, SOL and gastrocnemius, to quiet standing in healthy young and older adults. That study showed minimal lower limb muscle activity was needed to maintain quiet standing posture. Therefore, textured surfaces may have a greater effect on lower limb muscle activity in older people under conditions where substantial changes in muscle activation patterns occur in response to challenging, dynamic balance tasks.

Previous research suggests the inability to generate sufficient muscle force in the lower limbs can contribute to poor balance in older people (Horak et al., 1989). Ankle plantarflexor muscle force in older adults is a significant predictor of dynamic balance performance (Daubney and Culham, 1999). Measures of ankle plantarflexor strength and power have also shown to be significant predictors of dynamic functional performance in community-dwelling older females (Suzuki et al., 2001). Therefore, if textured surfaces can significantly reduce activity in lower leg muscles, this could have major implications on functional performance. Textured surfaces may provide sufficient sensory stimulus to alter neuromuscular mechanisms for balance control, reducing the extent of muscle activity needed for quiet standing.

9.4: Summary

The current study reported that textured surfaces can significantly alter CoP velocity and ML sway range in healthy older adults. This effect was shown when standing with eyes closed. Evidence suggests the geometric design specific to T1 may be capable of reducing sway parameters, and potentially improving balance. In comparison, T2 appears to have a de-stabilising effect on standing balance. Some older people may respond immediately to the application of plantar tactile stimulation, whilst others may require a longer adaptation or acclimatisation period. The rate at which enhanced plantar tactile stimulation is detected or processed could be dependent upon physical status.

Textured surfaces had no significant effects on lower limb muscle activity in healthy older adults. Changes in postural sway parameters may not occur as a result of alterations in leg muscle activation patterns. Five lower limb muscles were investigated: it is possible changes may have occurred in other muscles.

Exploratory sub-group analysis for gender indicated that textured surfaces may have a more marked effect on sway variables in healthy older females. Also, the two textured surfaces appeared to have significantly opposite effects on MG activity in older females. This speculative evidence points to the need for more detailed analysis into the gender-specific effects of texture on sensorimotor function in older people.

Quiet standing may have been an insufficiently demanding balance task to induce muscular responses for balance control. Traditional postural sway variables and measures of EMG amplitude may be too crude to detect differences between pairs of textured conditions. However, textured surfaces may be a useful intervention for improving balance in healthy older adults.

CHAPTER 10: GENERAL DISCUSSION AND CONCLUSIONS

10.1: Introduction

This chapter will evaluate the overall findings for the effect of textured surfaces on postural sway and lower limb EMG activity during quiet standing in healthy young and older adults. Differences in the sensorimotor effects of textured surfaces in young and older adults, and between conditions where visual information is available and suppressed, will be discussed. Consideration will be given as to how the CNS may use enhanced sensory information to control balance. Methodological strengths and limitations of the current study will be discussed and recommendations made for future research.

10.2: Aims of the Current Thesis

This thesis was concerned with investigating whether textured surfaces have a role to play in changing balance performance in healthy young and older adults, by providing enhanced plantar tactile stimulation. In chapter 2 of this thesis, the literature review indicated textured surfaces and textured foot insoles can significantly alter standing balance in healthy young adults (Corbin et al., 2007, Palluel et al., 2008, Watanabe and Okubo, 1981). It also identifed a significant lack of evidence relating to the effect of texture on balance performance in older adults: a clinical population known to present with postual instability and at high risk of falling. Only one previous study reports the potential for textured foot insoles to significantly improve standing balance in healthy older adults (Palluel et al., 2008). Therefore, the primary objective of the current thesis was to measure postural sway parameters during quiet standing on three different textured surfaces, with eyes open and eyes closed, over three different time intervals: the first 10, latter 20, and overall 30 seconds of quiet standing. By investigating three different textured surfaces, this thesis was designed in such a way that enabled generation of novel evidence relating to the importance of the geometric textured pattern in changing balance. To date, previous studies have investigated only one design of textured footwear intervention relative to a ‘no texture’ control condition in healthy young and older adults (Corbin et al., 2007, Kelleher et al., 2010, Nurse et al., 2005, Palluel et al., 2008, Watanabe and Okubo, 1981). By investigating standing balance with eyes open and eyes closed, this thesis had the scope to determine whether textured surfaces provide a source of plantar tactile stimulation which has the capacity to enhance balance when visual information is available, or compensate for a loss of balance due the removal of visual information when the eyes are closed. By investigating the effect of textured surfaces on postural sway and lower limb muscle activity over three different time intervals: the first 10, latter 20 and overall 30 seconds of quiet standing, this thesis was able to determine whether a textured intervention had greatest capacity to enhance balance during or following re-stabilisation of the body, after standing up from a seated position.

The current thesis also explored whether the contact area between the plantar surface of the foot and the textured surface in individuals with normal, high-arched and low-arched foot structure is a factor which influences the magnitude of the textured effect. However, this sub-group analysis was not possible as there were insufficient healthy young and older adults presenting high-arched and low-arched foot structure.

The literature review also indicated that textured insoles can significantly change lower limb muscle activity during walking in healthy young adults (Nurse et al., 2005) and adults with Multiple Sclerosis (Kelleher et al., 2010). One aspect that remained unclear was whether textured interventions would alter lower limb muscle activity in healthy older adults, as a component of sensorimotor function involved in upright balance control: this had not previously been explored in older people. Therefore, secondary aims of the current thesis were to investigate whether textured surfaces altered lower limb muscle activity during quiet standing in healthy older adults.

Therefore, this thesis provides novel evidence, addressing a number of important gaps in current literature including:

1. The effect of texture on postural sway parameters and lower limb muscle activity simultaneously during quiet standing with eyes open and eyes closed, over the first 10, latter 20 and overall 30 seconds, in healthy young and older adults

2. The importance of the geometric design of a textured surface in altering balance performance during quiet standing with eyes open and eyes closed, over the first 10, latter 20 and overall 30 seconds, in healthy young and older adults

10.3: Original Contributions to Scientific Knowledge

10.3.1: Postural sway

In healthy young adults the two textured surfaces appear to have opposite effects on AP sway during quiet standing with eyes open: however this finding did not reach the level of statistical significance. In healthy older adults, textured surfaces significantly altered CoP velocity and ML sway range during quiet standing with eyes closed. Speculative findings for the gender effects of texture on balance performance suggest textured surfaces may have a greater capacity to alter sway parameters in healthy older females, compared to older males.

This thesis points to the importance of the geometrical textured pattern of a surface for enhancing balance performance in healthy young and older adults. Relative to control, T1 reduced sway parameters. An opposite effect was seen with T2. This thesis suggests that a textured pattern similar to that of T1, comprising regularly spaced, small, pyramidal peaks, covering the full plantar aspect of the foot is capable of reducing postural sway in healthy adults.

In summary, textured surfaces significantly affect measures of ML sway and CoP velocity in healthy older, but only in the absence of visual information. Textured surfaces do not significantly alter postural sway parameters in healthy young adults, but there is tentative evidence suggesting textured surfaces may have some effect on AP sway, under conditions where vision is available.

Within the context of the current thesis, it is also important to compare and evaluate the effects of textured surfaces on postural sway and muscle activity during eyes open versus eyes closed test conditions.

Visual inputs are the primary source of sensory information used by the CNS for upright balance control, providing details of head position. In the absence of visual inputs, due to impairment or experimental manipulation, only vestibular inputs and proprioceptive receptors located in the neck can generate information about head orientation and position. Sensory information retrieved from the upper body is integrated with proprioceptive information below at the ankle joints. Accornero et al. (1997) propose that when visual sensory information is absent, sensorimotor control of upright balance is largely governed by the position of the joints and body segments, rather than the head. Therefore, postural control may become solely dependent on information from ankle proprioceptors, in combination with plantar mechanoreceptors. This may explain why changes in postural sway in older adults were only observed with eyes closed test conditions. Postural control may have been augmented by using textured surfaces to enhance somatosensory information (with eyes closed).

When visual inputs were available during quiet standing with eyes open, the textured surfaces may have provided excessive sensory stimulation, indeed over-stimulation, resulting in sensory conflict (Patel et al., 2009). The CNS may not have been capable of processing or responding to such enhanced stimulus, leading to no significant changes in balance performance. However, in healthy young adults, the trend for T1 and T2 to have opposite effects on AP sway were only observed in the latter 20 seconds and overall 30 seconds of quiet standing with eyes open. This finding suggests the importance of the textured geometric pattern may not become apparent until the body has re-stabilised, after standing up from a seated position, or adapted to a new sensory stimulus. Textured surfaces may only enhance balance control mechanisms where sensory conflict between visual, vestibular and somatosensory systems does not occur (Patel et al., 2009).

10.3.2: EMG

No significant changes were observed in the amplitude of lower limb muscle activity, between textured conditions in healthy young or older adults. However, when older adults stood on T1 there appeared to be a consistent pattern for decreased TA activity relative to the control surface. Speculative data for the gender effects of texture on lower limb muscle activity indicate that textured surfaces may be capable of altering MG activity in healthy older females.

It is important to consider why textured surfaces had no significant effects on lower limb muscle activity in healthy young adults or older adults. Should textured surfaces have significantly altered sway parameters in healthy young adults, it may be unlikely that such changes would have been accompanied by alterations in lower limb muscle activity. Previous research suggests healthy young adults have a greater reliance on cutaneous and proprioceptive sensory information for controlling upright stance, in preference to altering the intensity of lower limb muscle activity (Benjuya et al., 2004). Comparing tests of quiet standing with eyes open versus eyes closed, more pronounced increases in postural sway were observed in healthy young, compared to older adults, between conditions. However, there were no significant differences in simultaneous measures of lower limb muscle activity, between visual conditions in the younger adults (Benjuya et al., 2004). By increasing postural sway, younger adults may receive greater cutaneous sensory input from the feet and proprioceptive information from stretch receptors in the lower limb muscles. This sensory information may be sufficient to control or restore postural sway within safe limits, without the need to increase lower limb muscle activity.

Benjuya et al. (2004) reported that in healthy older adults, increases in postural sway between eyes open and eyes closed trials, were accompanied by significant increases in lower limb EMG activity, particularly TA and SOL. With increasing age, there is a shift from a reliance on sensory information to muscle co-contractions for maintaining upright standing balance (Benjuya et al., 2004). However, the current findings for healthy older adults do not support this postulation. In this thesis, when older adults stood on textured surfaces with their eyes closed, there were no significant changes in lower limb muscle activity. It is possible that a textured surface could assist in reversing the preference for older people to rely on muscular co-contractions for balance control. Enhancing plantar tactile stimulation using a textured surface may restore older peoples' reliance on sensory information, enabling them to use similar postural control mechanisms to healthy younger adults.

The specific original contributions to scientific knowledge that this thesis has generated, in comparison to previous research studies investigating texture, can be seen more clearly in Figure 10.1.

10.3.3: Textured surfaces and underlying mechanisms

The findings of the current thesis need to be considered within the context of how the CNS uses sensory information to control balance. It is not important how individual plantar mechanoreceptors interpret and transmit enhanced sensory information from the textured surfaces, but rather how the CNS uses this tactile stimulation to control standing posture. There are two key issues to explore: the direct mechanical effect and sensorimotor effect of textured surfaces.

The direct mechanical effect of textured surfaces relates to alterations observed in kinetic or kinematic data caused by a change in the location, magnitude and temporal patterns of the ground reaction force. These changes result from direct contact between the plantar surface of the feet and textured surface. In the current thesis, it is unlikely direct mechanical changes took place. The textured surfaces

	Investigated healthy young adults	Investigated healthy older adults	Textured surfaces	Textured foot insoles	Excluded confounding effects of footwear	2 or more different textured patterns	Quiet standing balance	Gait Analysis	Lower limb muscle activity	Measures of standing balance and EMG
Current thesis	√	√	√	X	√	√	√	X	√	√
Corbin et al. (2007)	√	X	X	√	X	X	√	X	X	X
Kelleher et al. (2010)	X	X	X	√	X	X	X	√	√	X
Nurse et al. (2008)	√	X	X	√	√	X	X	√	√	X
Palluel et al. (2008)	√	√	X	√	X	X	√	√	X	X
Watanabe and Okubo (1981)	√	X	√	X	√	√	√	X	X	X
Wilson et al. (2008)	√	X	X	√	X	√	√	√	X	X

Fig 10.1: Review of the original scientific contributions of the current thesis in comparison to previous research.

were not contoured to the plantar surface of the foot, so would not influence joint mechanics or biomechanical alignment. This thesis did not set out to explore the effect of textured surfaces on joint mechanics using motion analysis systems to generate biomechanical models, but rather provide a clinically relevant measure of balance performance in adults during quiet standing. Measures of postural sway, quantifying the distance and velocity of CoP movement, was the primary objective of this thesis, to determine whether textured surfaces may be a potential future intervention to counteract or restore balance impairments related to age, pathology or injury.

The sensorimotor effect of textured surfaces relates to observed alterations in kinetic or kinematic data caused by changes in sensory input to the CNS. This sensory stimulus can lead to changes in the rate or pattern of discharge of afferent information. This may have a subsequent effect on the magnitude and temporal pattern of efferent output, or motor stimuli, from the CNS to the muscles of the foot and lower limbs. The afferent inputs into the CNS that may lead to such changes may originate from sensory organs within the skin, joints, muscles, tendons, eyes, inner ears or other areas of the body.

It is most likely that the textured surfaces had a sensorimotor effect on healthy young and older people, by means of altering the rate of discharge or firing patterns from plantar mechanoreceptors. It was not within the scope of this thesis to explore sensory afferent activity, but it appears that the two different textured surfaces may affect this discharge in different ways. Changes in mechanoreceptor activity when standing on T1 showed to have favourable implications on standing balance in healthy older adults.

10.4: Main Conclusions

This research points to several main conclusions:

1. Textured surfaces may only significantly influence balance performance in older people: a population whose balance ability is known to deteriorate with increasing age and be poorer of that in younger adults.
2. Textured surfaces may have a greater effect on postural sway when visual sensory information is removed. This suggests texture may provide a

surrogate source of sensory information for balance control when one or more of the remaining sub-sensory systems are redundant.

3. Using texture, as a medium to enhance balance performance, may have a more marked effect in healthy older females, compared to males.

4. Changes in postural sway, resulting from enhanced sensory input at the feet, do not appear to be accompanied by significant underlying alterations in the amplitude of lower limb muscle activity. That is not to say significant changes in leg muscle activation patterns may not have occurred in other muscles not investigated, or in other measures of EMG activity.

5. There may be an optimal textured pattern, in terms of the geometric shape and distribution of indenting protrusions, which could be therapeutically beneficial for enhancing standing balance. It is possible this 'optimal' textured design may be similar to that of T1.

6. Adults may need to be exposed to a textured intervention for an optimal duration, before improvements in balance performance are observed.

7. Quiet standing balance appears to be an insufficiently demanding task to observe the effect of textured surfaces on postural sway and lower limb EMG measures in healthy adults.

10.5: Study Strengths

This thesis presents two robustly designed studies, using rigorous scientific methodologies that provide novel, clinically important findings which will inform future research. The first phase of this PhD, investigating healthy young adults, has already been published (Appendix 14) in a high quality, international journal. The findings from both healthy young and older adults have also been presented at national and international conferences, following peer-review selection (Appendices 31-35).

In the current thesis, the within-subjects research design was an efficient way of examining the effect of textured surfaces using the sample size of n=24 (healthy young adults) and n=50 (healthy older adults). The fact that the three surfaces were randomised reduced the possibility of bias linked with order effects.

The current thesis investigated the effect of textured surfaces rather than foot insoles. This was to prevent confounding effects of footwear including shoe construction, heel height, insole contouring and fastening mechanisms.

The current thesis adopted test procedures to minimise habituation to the textured stimulus. Participants had not been exposed to any of the textured conditions, prior to testing. Therefore, as the mechanoreceptors on the plantar surface of the feet were not habituated to this mode of tactile stimulus, it is presumed that observed postural and muscular responses to texture reflect the immediate response to a new source of plantar stimulation. Following each test, participants were asked to sit down and take their feet off the textured surfaces and force platform. This rest period served to prevent habituation to the sensory stimulus of the textured indentations.

10.6: Study Limitations

In the current thesis, neither the investigator nor participants were blind to the condition being tested. This procedure was not attempted as it was thought it would be obvious to the participants which texture they were standing on. That would certainly have been so with regard to the smooth, control versus textured surfaces. Participant blinding counters a bias in which the participants try to perform in line with what they think are the researchers' expectations. However, all of the information supplied to the participants emphasised the neutral stance of the researcher and the equipoise in the research question. Interaction between the researcher and participants was kept to a minimum set of instructions. No verbal encouragement was given before or during balance tests. Neither was there feedback on performance after the balance tests.

The current thesis investigated the effect of texture on postural sway during tests of quiet bipedal standing balance. This data is important, but cannot be extrapolated to more challenging balance conditions and functional movement. It is possible quiet standing was an insufficiently demanding balance test for healthy adults. However, within current literature there is no consensus as to which test of balance would be regarded the 'gold standard' to assess postural control in healthy adults.

10.7: Future Recommendations

This thesis has produced various important findings that require future investigation. Highlighted below are areas of necessary research that will help develop greater understanding of the potential for textured interventions to enhance functional performance in adults.

10.7.1: Balance tasks

The full extent of the effect of textured surfaces on balance performance in healthy adults may only become apparent when the postural control system is substantially challenged. Future research should set out to investigate whether textured surfaces can significantly affect postural stability under more challenging and vigorous dynamic balance tests such as single-limb stance, tandem stance, sit-to-stand, locomotion or dual-task conditions which introduce a secondary distraction task.

10.7.2: Outcome measures

In the current thesis, traditional CoP-based postural sway variables were used to analyse the effects of textured surfaces on balance. Measures of EMG amplitude were extracted to quantify the effect of texture on lower limb muscle activity. Future work should explore the effect of texture on non-traditional sway parameters, such as time-to-boundary, which expresses a different, yet related dimension of postural control in comparison to traditional sway measures. Time-to-boundary estimates the time it would take for the CoP to reach the boundary of the base of support, should the CoP continue on its path, at its current velocity (Hertel and Olmsted-Kramer, 2007). This measure is associated with spatiotemporal characteristics of postural control, taking into account the base of support formed by participant's feet, the instantaneous position and velocity of the CoP in AP or ML directions, as required (Hertel and Olmsted-Kramer, 2007).

A critical review relating to measures of postural stability should be undertaken to inform future analysis of texture and balance performance. There is currently no gold standard measurement technique, method of calculation or reporting of postural sway parameters. This lack of standardisation has been recognised by the International Society for Posture and Gait Research (ISPGR), who recently established a Standardization Committee for Clinical Stabilometry in order to create such standards. Therefore, the aforementioned critical review would address a major gap in current literature and follow in line with the future directions of the ISPGR.

It is also important to explore whether there are limits on reducing postural sway. Postural sway could be reduced to such an extent that a threshold is eventually reached, whereby no further reductions are observed, irrespective of continued exposure to a new, or the original intervention responsible for initial balance improvements. Without such evidence, it is difficult to determine the maximum potential of texture as an intervention for improving balance performance. These concepts require further investigation in healthy and pathological groups, at different time intervals over the course of a balance test, with eyes open and eyes closed.

Future research should also investigate how texture affects other components of the EMG signal such as wavelet frequency analysis, co-contraction patterns and onset timings. It may be advantageous to initially conduct a sophisticated and detailed analysis of how muscle activity contributes to postural control during various functional tasks. This would provide complimentary evidence, having evaluated the influence of textured conditions on leg muscle activity.

10.7.3: Study population

Future research is required to explore the effect of textured surfaces and footwear interventions on balance performance in frail older adults with a history of falling, injured and pathological groups such as rheumatoid arthritis, diabetes and neuromuscular diseases. The current thesis concluded that textured surfaces may have a greater effect on postural sway in the absence of visual sensory information. This would suggest texture may provide a surrogate source of sensory information for balance control when one or more of the remaining sub-sensory systems are redundant. Therefore, it would be advantageous to explore the effect of textured surfaces on balance performance in clinical populations with known sensory deficits such as adults with visual impairments including age-related macular degeneration, cataracts, glaucoma in diabetes, vestibular impairments, or peripheral sensory impairments such as neuropathy or hypersensitivity.

The importance of the findings from the current thesis has already been recognised by national UK funding bodies. The study investigating healthy older adults was part funded by the British Geriatrics Society. Thereafter, two further funding sources have supported research studies investigating:

1. The effect of textured insoles on postural stability and lower limb muscle activity in older adults prone to falling *(funded by the Physiotherapy Research Foundation).*

2. Wearing textured insoles in shoes to help self-management for people with Multiple Sclerosis: an exploratory study of the effects on balance *(funded by the Multiple Sclerosis Society).*

10.7.4: Further exploration of texture

Exploratory work is needed to establish whether there is an optimal geometric pattern including the prominence, centre-to-centre distance or distribution of protrusions over the plantar aspect of the foot. The current thesis provides evidence that there is a need to conduct detailed analysis to:

1. Discern the optimal exposure time to a textured intervention.

2. Understand how plantar tactile stimulation changes plantar cutaneous mechanoreceptor activity in the short and long-term, and whether habituation occurs.

3. Explore whether foot structure is a factor that influences the effect of textured surfaces on postural sway and lower limb muscle activity.

4. Explore whether textured surfaces, used as foot insoles, would bring about similar or greater effects on sensorimotor function in young and older adults, when introducing confounding effects of footwear.

5. Investigate the type of footwear used to deliver a textured insole. Palluel et al. (2008) investigated a textured sandal. Future research could investigate the effect of textured insoles within Oxford-type shoes for men, or moccasin-type shoes for women.

6. Explore the long-term effects of texture and the possibility of using textured insoles as a potential intervention to enhance balance in clinical populations, particularly those with visual, vestibular and peripheral sensory deficits.

10.8: Overall Conclusion

In conclusion, the purpose of this thesis was to explore the potential for different textured surfaces to improve balance performance in healthy adults, by way of providing enhanced plantar tactile stimulation. Two separate studies investigated the effect of textured surfaces on postural sway and lower limb muscle activity in healthy young and older adults, during quiet standing. Two different textured surfaces, and one surface as control, were investigated in order to determine whether an optimal geometric pattern may exist for enhancing balance performance. The current thesis also carried out exploratory sub-group analysis for gender effects of texture in healthy older adults.

Textured surfaces appear to have a significant effect on sensorimotor function in healthy adults. A textured surface comprising small, regularly spaced, pyramidal peaks, covering the whole plantar aspect of the foot can significantly improve postural sway parameters in healthy older adults: possibly to a greater extent in older females. Similar effects were tentatively observed in healthy young adults. This evidence is clinically important, suggesting a textured surface has the capacity to reduce the magnitude and velocity of CoP movement, thereafter reducing the demands placed on the postural control system for maintaining upright posture and the likelihood of experiencing loss of balance.

Textured surfaces do not significantly alter lower limb muscle activity in healthy young or older adults during quiet standing. This suggests changes in balance performance due to enhanced plantar tactile stimulation may not arise from underlying changes in the amplitude of lower limb muscle activity. However, there are speculative findings that textured surfaces may alter the intensity of ankle muscle activity, in particular the plantarflexors and dorsiflexors.

The effect of textured surfaces on postural sway and lower limb muscle activity in adults may be augmented during more challenging balance tasks, in clinical populations, or upon analysing a wider range of postural sway and EMG outcome measures. There appears to be an optimal geometric pattern for enhancing balance performance. The implications for clinical practice are that texture could be incorporated into interventions designed to enhance balance, such as foot insoles.
Further investigation is necessary.

Effect of foot orthoses on lower limb muscle activation: a critical review

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Foot orthoses can be a valuable component of musculoskeletal rehabilitation, improving lower limb alignment, controlling motion and providing shock absorbency. Recent evidence suggests foot orthoses may also have a significant effect on lower limb muscle activation in young, healthy adults. This review examines the evidence for changes in muscle activation patterns when wearing orthoses, and explores the proposed mechanisms by which foot orthoses may bring about changes in lower limb muscle activity. Based on the current results it is proposed that different mechanisms may occur by which orthoses affect muscle activity, due to their differing construction and design.

Keywords: electromyography, EMG, lower limbs, muscle activity

Introduction

Health care professionals frequently prescribe foot orthoses (FOs) as a treatment modality for the treatment and prevention of overuse running injuries. A review of the literature demonstrates a wide-range of overuse lower limb injuries for which FOs have been used as a form of intervention, including Achilles tendonitis,[1] iliotibial band syndrome,[2] patellofemoral pain syndrome,[3] plantar fasciitis,[4] posterior-tibial tendon dysfunction[5] and medial tibial stress syndrome.[6] FOs have been reported to increase comfort and improve proprioception in both the athletic and high-risk groups.[7–10]

The exact biomechanical mechanism of action of FOs remains unknown. However, FOs are often prescribed to improve lower extremity alignment by means of controlling abnormal or excessive movement at the subtalar joint. In contrast, some authors have suggested that the shock-absorbing effects of orthoses, rather than their ability to correct alignment and control motion, may be their most useful asset.[7,8]

To improve lower extremity alignment, mechanical, neurophysiological and anatomical factors must be addressed. Mechanical factors can affect lower limb biomechanical alignment as foot movement is transferred to the tibia via a coupling mechanism. Therefore excessive inversion/eversion at the subtalar joint can be translated to increased external/internal tibial axial rotation.[11] Orthotic support at the medial arch of the foot is considered a standard corrective intervention for excess foot pronation and has been shown to place the foot in a more inverted position during the stance phase of running,[12,13] producing a mean reduction in the range of internal tibial rotation during running,[14] and maximum calcaneal eversion during fast walking.[15]

Similarly, there is evidence that neurophysiological factors can affect lower limb biomechanical alignment, suggesting that disrupting the function of intrinsic foot muscles, via induced fatigue, results in increased foot pronation,[16] and that impact forces during daily physical activity produce a reaction in the muscles to minimise soft tissue vibration.[17] Foot orthoses can be administered to augment the medial longitudinal arch, thus altering the muscular work required to maintain foot posture, and may act to dissipate and absorb this impact, in turn affecting levels of fatigue, intensity of muscular work, performance and comfort.[17] FOs may also provide

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biofeedback when placed under the mid-foot and forefoot, by enhancing afferent feedback from cutaneous receptors on the plantar surface of the feet and reducing eversion due to contraction of the inverting muscles.[12]

In addition, anatomical factors can affect lower limb biomechanical alignment: induced excessive foot pronation has been shown to significantly increase internal tibial rotation, internal hip rotation, and anterior pelvic tilt.[18] Correcting anatomical malalignment at the foot using orthoses may indirectly re-align the lower limb more superiorly.

Nigg *et al.* postulated that by inserting a corrective interface between the plantar aspect of the foot and sole of a shoe, the FO can counteract abnormal biomechanics, support natural joint motion and alter muscle activation patterns. Information generated by the ground reaction force is filtered by the FO and transferred to the central nervous system, which elicits a subject-specific dynamic response.[7]

Electromyography (EMG) measures the electrical activity associated with muscle activation and is used in gait analysis with a simultaneously obtained signal from an electronic walkway or a series of footswitches, to identify phases of the gait cycle. Monitoring EMG activity during gait can play an important role in evaluating walking and running performance in individuals with lower limb overuse running injuries, and can also be used to establish key training parameters for facilitating optimal running patterns.

It has been proposed that for any given movement task, the skeleton has a preferred path. The locomotor system may choose a strategy to keep skeletal movement in a constant path of minimal joint resistance.[19] A shoe, insert or orthotic intervention can support or counteract the preferred movement path, causing the corresponding muscle activity to be reduced or increased.[7] Deviation from the movement path of minimal resistance may trigger a neuromuscular response, with the appropriate muscles being activated to oppose any intervention, thus causing an alternative path of skeletal movement.[17] Therefore, any intervention that supports this preferred path may provide greater comfort, and reduce muscle activity.[7] When the foot pronates more than normal, the tibialis posterior tendon pulls harder at its insertion on the navicular in order to supinate the foot during the gait cycle. Medial wedging will mechanically increase the medial longitudinal arch of the foot, altering the pull of the tibialis posterior tendon, reducing muscle activity, and thereafter reducing the potential for tendonitis. Likewise, tibialis anterior, one of the major inverter muscles of the foot, compensates for overpronation and is susceptible to tendonitis. Functional FOs providing medial support can offload the medial plantar aspect of the foot, preventing tibialis anterior overuse symptoms.[20] In excessive foot pronation, orthotic management to place the heel in a more vertical position will prevent pinching of the peroneal tendons between the lateral aspect of the calcaneus and tip of the fibula, which can lead to peroneal tendonitis. Re-aligning the heel can also reduce the duration of activity in the gastrocnemius-soleus muscle complex, with such prolonged contraction being common in Achilles tendonitis.[20]

Current evidence from the literature suggests that FOs may affect lower limb muscle activity while walking and participating in physical activities.[21–30] However, there is a wide array of FOs available and use is based on personal choice, materials available and costs, rather than on claims regarding their function.[31] Additionally there is limited knowledge about the specific mechanisms by which FOs influence EMG activity.

Therefore, the aim of this review was to assess the evidence for how FOs can influence lower extremity muscle activation, and to review proposed theories for the underlying mechanisms by which such changes occur.

Methodology

Literature searches of AMED (1985 to October 2006), Medline (1966 to October 2006), CINAHL (1982 to October 2006), PEDro, EMBASE (1980 to December 2006) and the Cochrane Database, in addition to hand-searches and references from journal articles, were undertaken between 8 May 2006 and 1 December 2006. The language was restricted to English and the following terms were used in the search strategy: 'foot orthoses', 'feet', 'orthotics', 'muscle activity', 'insoles', 'inserts', 'textured insoles', 'surfaces', 'afferent input', 'stochastic resonance', 'wedging', 'lower limbs', 'lower extremities', 'electromyography', 'somatosensation', 'plantar surface', 'cutaneous receptors', 'balance', 'physical performance', 'sensory input', 'postural control', 'gait'.

The term FO includes many different types of orthotics: flexible, semi-rigid, biomechanical, insoles and shoe inserts. For the purposes of this review the terms rigid, semi-rigid, and soft foot orthoses were used to describe the variety of FOs available and the impact on EMG activity, based on materials used in their construction.[31] Although not strictly described as an FO, therapeutic footwear was included in the review.

Results

The electronic search initially generated 27 hits. To be included in the review, a trial had to meet both of the following inclusion criteria:

- Participants received a foot orthosis or therapeutic footwear.
- Outcome measures included analysis of lower limb muscle activation.

15 papers were excluded in total: one paper did not meet either inclusion criterion, as it investigated the effect of shoe midsole hardness on dynamic balance control. A further 14 papers only met one of the inclusion criteria and were excluded from the review; of these, four analysed the effect of different shoes or shoe midsole characteristics on lower limb EMG. The remaining 10 papers evaluated the effect of foot orthoses on other outcome measures including gait kinematics, vertical loading, postural sway and joint movement discrimination (Fig. 1).

One reviewer undertook the searches and assessed potential studies against the inclusion criteria (ALH). Papers reviewed in this article are summarised in Table 1. Meta-analysis was not possible as the studies found were not randomised controlled trials.

The STROBE[32] checklist was used to rate the following aspects of methodological quality: rationale, objectives, study design, statistical methods, bias, data analysis, key findings, limitations, external validity and interpretation. The checklist consists of 22 items for consideration relating to cohort, case-control and cross-sectional studies. All 22 items were applied to each study reviewed and scored using the following scale: yes=1, unable to determine=0, no=0.[33] Table 2 presents the results of the methodological quality assessment of included studies. Quality assessment results were expressed as a percentage, with scores ranging from 50 to 86% (mean=76·8%). Seven papers scored above 80%.

An adequate description of the study rationale and background was given in most studies, however six studies failed to describe the research setting and location. Rationale for sample size was rarely addressed, with 10 papers providing no reference to statistical consideration for the number of participants recruited. Lower scoring papers failed to provide sufficient detail concerning sources of

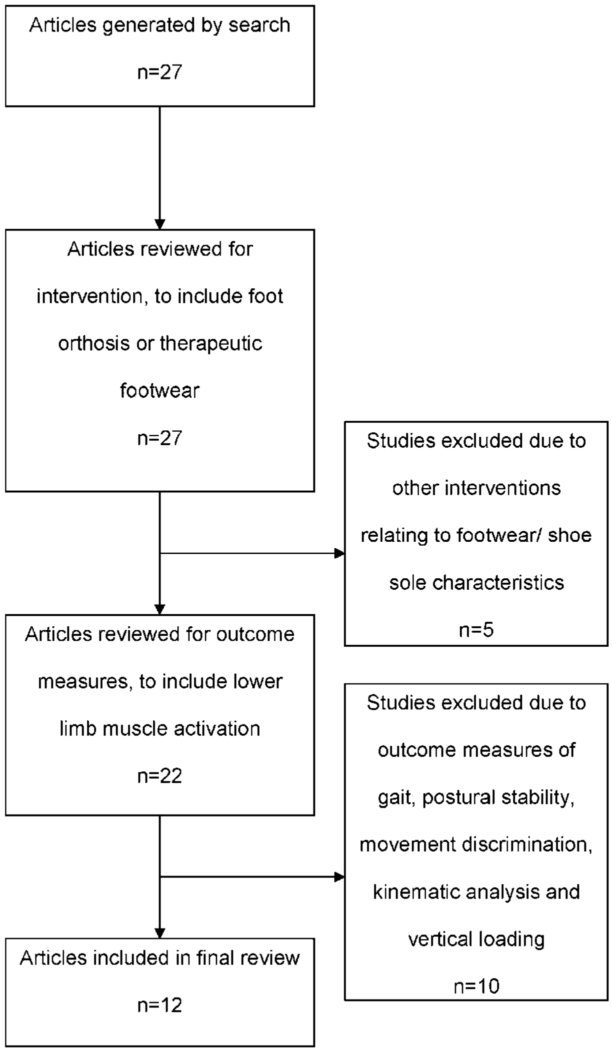

Figure 1 Flow chart of literature search for foot orthoses and lower limb muscle activation

funding, reporting of the number of participants during each stage of the study, or reasons for non-participation. Within the main results, seven papers did not provide accompanying statistical data to confirm the precision of their findings (e.g. 95% confidence intervals). Other methods of analysis, including subgroup analysis, were poorly reported, although it is acknowledged that this was not always possible or necessary within the respective papers. Five papers failed to adequately interpret the external validity or generalisability of their findings.

The papers reviewed were grouped into four categories; rigid, semi-rigid, soft and therapeutic footwear (Tables 3–6).

Semi-rigid foot orthoses (Table 3)

Nawoczenski and Ludewig[23] evaluated the effects of semi-rigid FOs on 12 subjects presenting with low arched feet and at least one month's history of

Table 1 Summary of papers reviewed

Author	Subjects	Age ± SD (yrs)	EMG	EMG analysis	Type of orthoses	Footwear
Tomaro and Burdett[26]	10 (7F, 3M)	25-30	Tibialis Anterior Peroneus Longus Lateral Gastrocnemius	EMG root mean square during walking.	Rigid custom-made foot orthotic.	Athletic Shoe.
Nawoczenski and Ludewig[23]	12 (6F, 6M)	27·2 ± 9·9	Tibialis Anterior Medial Gastrocnemius Vastus Lateralis Vastus Medialis Biceps Femoris	Root mean square amplitude for the first 50% of running stance phase.	Semi-rigid 3·2 mm polypropylene. Extrinsically posted in rearfoot and intrinsically posted in forefoot. Covered with 3·2 mm polyurethane.	Standardised Teva running sandal, with 25 mm heel height and strap design.
Rose *et al.*[29]	17 (4F, 13M)	20·6 ± 1·8	Medial Gastrocnemius Lateral Gastrocnemius Medial Hamstrings Lateral Hamstrings Medial Quadriceps Lateral Quadriceps	Muscle response time: from the onset of dominant lower limb perturbation, to the onset of long latency reflex. Activation order.	Semi-rigid custom moulded, full-length sport orthotic device.	Self-selected athletic shoe.
Bird *et al.*[30]	13 (7F, 6M)	22·3 ± 3·4	Erector Spinae (L3 level) Gluteus Medius	Time of onset and maximum amplitude of EMG activity during walking.	Rigid high-density ethyl vinyl acetate foot wedges, with 5° lateral forefoot wedge, 5° medial forefoot wedge and heel lift with maximum height 2 cm.	Not applicable.
Vanicek *et al.*[27]	6 (1F, 5M)	29 ± 10·5	Vastus Lateralis	EMG median frequency of the power density spectrum, during simulated ski squats.	Full-length closed-cell foam provides cushioning, with 3/4-length polypropylene heel caps. Green foot orthoses designed for high volume footwear and blue foot orthoses for low volume footwear (based upon volume of foam)	Ski Boots.
Hertel *et al.*[22]	30 (15F, 15M)	21·1 ± 1·6	Vastus Medialis Vastus Lateralis Gluteus Medius	Maximum root mean square during functional tasks.	Rigid off-the-shelf foot orthoses. Shell containing 7° medial and 4° lateral rearfoot posting.	Subject's own low-top athletic shoes.
Kulig *et al.*[24]	6	25·0 ± 2·0	Magnetic resonance imaging of: Tibialis Posterior Tibialis Anterior Soleus Medial Gastrocnemius Peroneus Longus	Magnetic resonance imaging following (<3 min) resisted foot adduction with plantarflexion exercise.	Semi-rigid full-length orthotic with rearfoot and midfoot control, a patented support bridge, shock absorption system.	Shoes.
Nurse *et al.*[28]	15 (3F, 12M)	24·7 ± 2·9	Soleus Tibialis Anterior Medial Gastrocnemius Vastus Medialis Rectus Femoris Biceps Femoris	Frequency content analysis using wavelet technique for low and high frequencies during walking.	Semi-rigid 3 mm thick ethyl vinyl acetate foam insert, cut from commercial sandal. Textured with semi-circular mounds with centre distances of 8 mm.	Not applicable.

Table 1 Continued

Author	Subjects	Age ± SD (yrs)	EMG	EMG analysis	Type of orthoses	Footwear
Mündermann *et al.*[21]	21 (12F, 9M)	27·4 ± 1·8	Tibialis Anterior Peroneus Longus Medial Gastrocnemius Biceps Femoris Vastus Lateralis Vastus Medialis Rectus Femoris	Wavelet analysis during running for: - Overall EMG intensity - EMG intensity in the high frequency band - EMG intensity in the low frequency band	Rigid posted 6 mm full-length ethylene vinyl acetate wedge. Rigid custom-moulded polypropylene shell (unknown thickness). Rigid posted and moulded wedges. As for custom-moulded but with 6 mm extrinsic ethylene vinyl acetate post added to medial rearfoot and forefoot areas of polypropylene shell.	Standardised running sandals.
Murley and Bird[25]	17 (10F, 7M)	23 ± 5	Tibialis Anterior Soleus Peroneus Longus Medial Gastrocnemius	Onset and maximum EMG amplitude during walking.	Rigid custom-made foot orthoses with medial rearfoot wedging at 0°, 15° and 30°.	Standardised shoes with canvas upper and flat thin rubber sole.
Nigg *et al.*[40]	8 (3F, 5M)	28·0 ± 3·6	Tibialis Anterior Medial Gastrocnemius Biceps Femoris Vastus Medialis Gluteus Medius	EMG intensity using Wavelet Analysis Technique during walking.	Not applicable	Masai Barefoot Technology Shoe. Control Shoe: Adidas SuperNova running shoe.
Romkes *et al.*[39]	12 (6F, 6M)	38·6 ± 13·2	Medial Gastrocnemius Lateral Gastrocnemius Tibialis Anterior Vastus Medialis Vastus Lateralis Rectus Femoris Semitendinosus	Root mean square for 16 time intervals of walking gait cycle.	Not applicable	Masai Barefoot Technology Shoe.

activity-related lower extremity musculoskeletal overuse symptoms (Table 3). Results demonstrated a statistically significant decrease of 11·1% in the average EMG amplitude of biceps femoris activity and a statistically significant increase of 37·5% in tibialis anterior activity while running with a sandal including the semi-rigid FO. Mean decreases in EMG activity in vastus medialis, vastus lateralis and medial gastrocnemius were not significantly different from the non-FO condition of sandal only. Rose *et al.*[29] demonstrated that the short-term application of semi-rigid FOs does not significantly affect reactive

Table 2 STROBE (strengthening the reporting of observational studies in epidemiology)*

	Item no.																						
Paper	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22	%
Tomaro and Burdett[26]	1	1	1	1	0	1	1	1	1	0	1	1	0	0	1	1	0	0	1	1	0	1	68
Nawoczenski and Ludewig[23]	1	1	1	1	0	1	0	1	1	0	1	1	0	0	1	1	0	0	1	1	1	1	68
Rose *et al.*[29]	1	1	1	1	1	1	1	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	82
Bird *et al.*[30]	1	1	1	1	1	1	1	1	0	0	1	1	0	0	1	1	1	1	1	1	1	1	82
Vanicek *et al.*[27]	1	1	1	1	0	1	1	1	1	0	1	1	0	1	1	1	1	1	1	1	1	1	86
Hertel *et al.*[22]	1	1	1	1	1	1	1	1	1	0	1	1	0	0	1	1	1	0	1	1	1	1	82
Kulig *et al.*[24]	1	1	1	1	1	1	1	1	1	0	1	1	0	0	1	1	0	1	1	1	1	1	82
Nurse *et al.*[28]	1	1	1	1	1	1	1	1	0	0	1	1	1	0	1	1	1	1	1	1	1	1	86
Mündermann *et al.*[21]	1	1	1	1	1	1	1	1	1	0	1	1	1	0	1	1	1	1	1	1	0	1	86
Murley and Bird[25]	1	1	1	1	0	1	1	1	1	0	1	1	0	1	1	1	0	0	1	1	0	1	73
Nigg *et al.*[40]	1	1	1	1	0	1	1	1	1	1	1	1	1	0	1	1	0	0	1	1	0	1	77
Romkes *et al.*[39]	1	1	0	1	0	0	1	1	0	0	1	0	0	0	1	1	0	0	1	1	0	1	50

*All items were scored using the following scale: yes=1, unable to determine=0, no=0

Table 3 Summary of papers investigating semi-rigid foot orthoses

	Semi-rigid FO			
Paper	Total subjects	Demographics of subjects	Quality score (/22)	Quality score (%)
Nawoczenski and Ludewig[23]	12	Symptomatic for lower extremity pain. Structural malalignment of the foot confirmed radiographically and clinically	15	68
Rose *et al.*[29]	17	Healthy, physically active, presenting hyperpronation of the subtalar joint >10 mm	18	82
Vanicek *et al.*[27]	6	Healthy alpine skiers	19	86
Kulig *et al.*[24]	6	Healthy, with an arch index of at least 2 standard deviations below normative values	18	82

Table 4 Summary of papers investigating rigid foot orthoses

	Rigid FO			
Paper	Total subjects	Demographics of subjects	Quality score (/22)	Quality score (%)
Tomaro and Burdett[26]	10	History of leg pathology, resulting from compensatory subtalar joint pronation that had been treated with orthotics for a minimum of six months before participation in study and reported an improvement	15	68
Bird *et al.*[30]	13	Healthy, right-handed	18	82
Hertel *et al.*[22]	30	Healthy, recreationally active. Architectual foot type: pes cavus n=5 (F), N=5 (M), pes planus n=5 (F), N=5 (M), pes rectus, n=5 (F), N=5 (M)	18	82
Mündermann *et al.*[21]	21	Healthy, recreational runners, presenting >13 foot eversion when running at 4 m s^{-1}	19	86
Murley and Bird[25]	17	Healthy, with pronated feet, scoring >+6 on the Foot Posture Index	16	73

Table 5 Summary of papers investigating soft foot orthoses

	Soft FO			
Paper	Total subjects	Demographics of subjects	Quality score(/22)	Quality score (%)
Nurse *et al.*[28]	15	Healthy, asymptomatic	19	86

intentional or reflexive muscular response times, or activation order of lower limb muscles responsible for stabilising the knee when two different perturbations, internal rotation and external rotation, are applied to the dominant leg (Table 1). Subjects presenting with excessive flatfeet showed no significant difference in overall muscle response times between FO and non-FO conditions for either direction of perturbation.

Kulig *et al.*[24] identified improved selective recruitment and activation of tibialis posterior immediately following resisted foot adduction and plantarflexion exercises in healthy subjects with asymptomatic flatfeet, wearing an FO and shoe, over and above barefoot conditions (Table 1). Selective activation referred to the relative increase in the activity of tibialis posterior during closed chain resisted foot adduction compared with corresponding activity in gastrocnemius, tibialis anterior, soleus and peroneus longus. The opposite would therefore be assumed for non-selective activation.

Tibialis posterior was the only muscle showing a consistent increase in signal intensity after both barefoot and FO exercise conditions.[24] However, average increase in signal intensity in tibialis posterior nearly doubled when exercising with the FO (54%) compared to the baseline, barefoot condition (29%, $p=0{\cdot}019$). The magnitude of change with the FO in tibialis posterior activity was greater than that of all the remaining muscles, all of which failed to reach the 10% significance threshold during the resistive adduction exercise.

Measures of transverse relaxation times (T2s) after exercise, obtained using magnetic resonance imaging, show strong correlation to EMG recordings during resistive exercises.[34] The T2s of water in active, healthy, skeletal muscle, have been shown to increase during exercise and return to resting values one hour post-exercise,[35] with the rate of T2 change increasing linearly with greater muscle work rate.[36] Therefore, a muscle was defined as being active during exercise if its post-exercise signal intensity exceeded the baseline level by 10%.[24]

Kulig *et al.*[24] suggested that the greater activation of tibialis posterior with an FO, may indicate that a low arch, as in pes planus, could elicit dysfunction in tibialis posterior activation. Tibialis posterior acts to stabilise the foot during ambulation and its tendon supports the longitudinal arch of the foot. Dysfunction in tibialis posterior may disrupt controlled foot pronation and shock absorption during gait, causing ‘locking’ of the tarsal joints in terminal stance, and can contribute to acquired flat foot. Kulig *et al.*[24] demonstrated that an FO can mechanically support the arches of the foot and enhance selective activation in tibialis posterior in adults with pes planus to a level equivalent to those with a normal arch index, when performing the exercise barefoot.[24]

Rigid foot orthoses (Table 4)

Murley and Bird[25] investigated the effects of rigid custom-made FOs in 17 asymptomatic subjects, with low arched feet (Table 1). A statistically significant increase was observed in the mean percentage of maximum EMG amplitude of tibialis anterior with shoes only (30% increase), and at 0° (33% increase), 15° (38% increase) and 30° (30% increase) of FO medial wedging in contrast to the barefoot baseline. However, there was no significant graded response observed between the shoe and all three corrective FOs or between individual FOs. No significant changes in EMG maximum amplitude were observed for medial gastrocnemius and soleus. A statistically significant increase in maximum EMG amplitude (19% increase) was observed in peroneus longus with

Table 6 Summary of papers investigating therapeutic footwear

	Therapeutic footwear (Masai Barefoot Technology Shoes)			
Paper	Total subjects	Demographics of subjects	Quality score (/22)	Quality score (%)
Nigg *et al.*[40]	8	Healthy, asymptomatic	17	77
Romkes *et al.*[39]	12	Healthy, asymptomatic	11	50

15° medial wedging in contrast to the barefoot baseline. Although a general trend for increased peroneus longus amplitude was reported, this change was neither significant nor graded between the three levels of medial wedging, as progressive increases in amplitude were not observed from 0 to 30° of wedging. Murley and Bird[25] propose this may indicate that alterations in peroneal EMG amplitude are attributable to contact between an FO and the talonavicular arch region of the foot, rather than to the degree of rearfoot wedging.

Hertel *et al.*[22] studied the short-term effect of rigid FOs with four types of intrinsic rearfoot postings during functional activities (single-leg squat, lateral step-down and maximum vertical jump) in 30 subjects presenting pre-existing foot pathologies (Table 1). Hertel *et al.*[22] demonstrated that EMG activity in vastus medialis and gluteus medius significantly increased during the single-leg squat in all FO conditions when compared to the non-FO condition of shoes only. During lateral step-down, activity in vastus medialis was also significantly higher with any FO than in the non-orthotic condition. No significant change in muscle activity was observed in gluteus medius during the lateral step-down. Vastus lateralis failed to show any difference in EMG activity in either single-leg squat or lateral step down with FO wear. However, a significant decrease in vastus lateralis activity was observed during the maximum vertical jump when FOs were inserted into the shoe. No single orthotic was observed to be more advantageous than another for any specific muscle or activity tested.

Mündermann *et al.*[21] investigated the effects of custom-made FOs on lower extremity muscle activity in healthy recreational runners with flatfeet (Table 4). Of all the muscles evaluated, tibialis anterior, peroneus longus and biceps femoris generally showed the highest increases in overall EMG intensity when FO were worn. Rigid posted FOs produced the highest increases in overall EMG intensity in peroneus longus and biceps femoris at pre-heel-strike (50 ms before heel-strike) and post-heel-strike (50 ms after heel-strike). At these specific time points, the highest increases in overall EMG intensity observed with the rigid moulded FO, without extrinsic posting, were again found in peroneus longus, biceps femoris and also medial gastrocnemius. Rigid, posted and moulded FO brought about the highest changes in overall EMG intensity in both vastii muscles and rectus femoris during the remaining 30–100% of stance phase following pre-heel-strike. Mündermann *et al.*[21] concluded that the effects of posting and custom-made FOs on EMG intensity differed between the three phases of running gait, and were specific to the muscles and FO conditions.

Tomaro and Burdett[26] found that healthy adults treated with FOs demonstrated no statistically significant difference in average EMG muscle activity of selective lower limb muscles when walking in footwear with rigid FO, compared to footwear only (Table 1). These results are not in agreement with Mündermann *et al.*[21] who observed significant changes in tibialis anterior, peroneus longus and medial gastrocnemius activity with rigid orthoses while walking. However, these studies cannot be directly compared; although a similar construction of FO was administered and similar muscles investigated, their respective procedures for data acquisition and extraction differed: Tomaro and Burdett measured average EMG activity and duration of muscular activity,[26] while Mündermann *et al.* used wavelet analysis techniques.[21] However, a significantly longer duration of tibialis anterior activity following heel-strike was demonstrated with the FO condition.[26] Tomaro and Burdett[26] speculate this may reflect the role of orthoses in decelerating subtalar joint pronation velocity; a greater duration of time may be required to allow the forefoot to contact the ground in the FO condition. As tibialis anterior controls loading of the forefoot in stance this would substantiate its longer duration of activity.

Bird *et al.*[30] demonstrated that heel lifts and bilateral lateral wedging made from high-density ethyl-vinyl-acetate provoked a significantly earlier onset of muscle activity in erector spinae at ipsilateral heel lift during gait, i.e. weight acceptance of the contra-lateral limb, relative to the barefoot condition in healthy subjects (Table 1). Medial wedging showed no significant difference in the onset time of erector spinae. Both bilateral and unilateral heel lifts on the ipsilateral side generated a significant delay in the onset of gluteus medius activity at ipsilateral heel contact, in contrast to the barefoot condition. No significant differences in EMG amplitude were observed for either gluteus medius or erector spinae with any variant of wedging tested. Bird *et al.*[30] suggest that although earlier onset of activity of erector spinae and delayed onset of gluteus medius with varying foot wedges were minimal, these effects may be clinically significant because of the number of times the gait cycle is repeated within an average person's daily life, and the potential for timing-deficits in muscle activation to contribute to back pain.

Vanicek *et al.*[27] investigated the effect of high and low volume semi-rigid FOs on the EMG fatigue response of vastus lateralis during a sustained ski squat in six healthy alpine skiers (Table 1). Both the FO condition and the control condition of ski boot only showed significant differences in the EMG median frequency value during the first and last 20 s of the squat, signifying the occurrence of muscle fatigue. However, the high volume FOs demonstrated significantly less reduction in EMG median frequency within the final 20 s, compared to the low volume FOs and control. EMG decline between the first and last 20 s of the squat was not found to be significantly different between the low volume FO and control condition.

Soft foot orthoses (Table 5)

Nurse *et al.*[28] demonstrated a significant decrease in the overall EMG intensity of soleus and tibialis anterior for the entire stance phase with the textured FO, relative to the smooth 3 mm ethyl-vinyl-acetate foam control condition. Analysing the mean EMG amplitude within specific frequency bands demonstrated reductions in soleus and tibialis anterior intensity only within the low frequency spectrum; only soleus reached a level of significance. No other significant differences were observed for any other muscle or time interval. As the results from Nurse *et al.*[28] are specific to the low-frequency EMG band, this may suggest that the sensory effects of textured FOs were primarily influential on slower motor units,[37] given that their slower conduction velocity would lower the frequency content of the EMG signal.[38] However, during steady, level-ground walking it would be expected that slow twitch fatigue-resistant muscle fibres would be recruited and consequently it is possible that textured FOs may have affected slower motor units purely due to their predominance in the nature of the task.

Mündermann *et al.*[21] used similar EMG intensity thresholds to Nurse *et al.*,[28] but found greater relative increases in the high-frequency band while running with custom moulded and posted FOs. This may support the notion that changes to muscle activation by altered sensory input on the plantar aspect of the foot are influenced by the intensity of the activity performed. Excluding the mechanical component of FOs and the compounding effects of footwear, Nurse *et al.*[28] altered only the texture of the FO, which was stuck directly to the plantar surface of subjects' bare feet. Therefore, the results indicate changing the texture rather than the geometry of FO may influence lower limb muscle activity, via alterations in plantar sensory feedback.

Therapeutic footwear (Table 6)

Romkes *et al.*[39] reported decreased tibialis anterior activity during initial foot contact and loading, increased tibialis anterior activity during the entire swing phase and increased medial and lateral gastrocnemius activity from terminal swing to midstance, during ambulation trials with Masai Barefoot Technology Shoes in healthy subjects (Table 1). Vastus lateralis and vastus medialis responded to the therapeutic shoes with elevated activity from mid-stance to toe-off, while rectus femoris activity increased in mid-stance and reduced in the stance-to-swing transition. The Masai Barefoot Technology Shoe is a therapeutic intervention designed to strengthen the lower limb musculature during ambulation. The rounded soft sole in an anterior-posterior direction underneath the heel and rocker bottom provides an uneven surface, which simulates barefoot walking action, therefore challenging the muscles to be more active.[39] Nigg *et al.*[40] also investigated the effect of the Masai Barefoot Technology Shoe on several lower limb muscles during bilateral standing tests and walking trials in young healthy adults (Table 1). In comparison to a standardised control shoe, only tibialis anterior demonstrated a significant increase in EMG intensity with the Masai Barefoot Technology Shoes during the bilateral standing test. This change in muscle activity accompanied a significantly greater centre of pressure excursion during quiet standing in the unstable, therapeutic shoes. Following multiple walking trials, Nigg *et al.*[40] reported no significant differences in EMG intensity for any muscle when the overall EMG intensity for the muscles of interest was normalised to levels for the control shoe.

Discussion

This review demonstrates that a wide variety of FOs, with rigid, semi-rigid and soft construction, and therapeutic footwear have been investigated for their effects on lower limb muscle activity during static and dynamic activities (Table 1). While it is acknowledged that statistically significant changes in muscle activity, whether increased or decreased, may neither reach levels of clinical significance nor bring about therapeutic or functional benefit, some clear trends were seen in the studies.

There is some evidence to suggest that tibialis anterior activity while walking is reduced in the presence of FOs. Both Nurse *et al.*[28] and Romkes *et al.*[39] reported a reduction in tibialis anterior activity during gait trials while wearing textured shoe inserts and therapeutic shoes, respectively. Previous findings from Tomaro and Burdett[26] concluded that semi-rigid FOs caused an increase in the duration of tibialis activity when walking. This evidence may complement that of decreases in muscle activity observed by Nurse *et al.*[28] and Romkes *et al.*,[39] suggesting that a lower level of muscle activity may equate to greater fatigue resistance.

Only Tomaro and Burdett[26] investigated participants who presented with pronated feet and a history of lower limb injury. However the similarities with the findings of Nurse *et al.*[28] and Romkes *et al.*[39] may suggest FOs are capable of decreasing the muscle force needed to resist foot pronation during the first half of the stance phase in individuals with both normal and abnormal foot positioning.[41]

As the muscles around the ankle significantly contribute to the overall moment around the ankle mortise, reduced rearfoot inversion moment and negative work may indicate decreased loading of the extrinsic foot muscles controlling eversion, which include tibialis anterior, tibialis posterior, gastrocnemius and soleus. In light of this group of extrinsic foot muscles, it is relevant that Nurse *et al.*[28] also reported decreased activity in soleus, while Romkes *et al.*[39] and Mündermann *et al.*[21] both observed increased gastrocnemius activity, collectively showing that FOs brought about alterations in the level of activity in all the primary foot inverter muscles. Inverted orthoses may therefore reduce the work done by the rearfoot inverters, in those with and without excessive foot pronation.[41]

In accordance with Tomaro and Burdett,[26] Vanicek *et al.*[27] provide further evidence for the effect of FOs in reducing muscular fatigue in vastus lateralis, which could essentially lead to increased duration of muscle activity. Although their results are specific to vastus lateralis during a functional squat exercise, both studies indicate the effect of FOs on neurophysiological factors in terms of fatigue-resistance. It has also been shown elsewhere that custom foot orthoses can reduce the negative effects of lower limb muscle fatigue on postural stability.[42]

Tomaro and Burdett[26] investigated FOs providing a varus correction of excessive rearfoot pronation, whereas the FOs and therapeutic footwear used by Nurse *et al.*[28] and Romkes *et al.*[39] were interventions reliant on a textured surface and rounded, unstable, soft sole, respectively. This may suggest that the geometrical properties of FOs are not solely responsible for eliciting changes in lower limb muscle activity.

In contrast to the findings of reduced tibialis anterior activity,[26,28,39] several papers provide evidence for increased tibialis anterior activity with FOs[21,23] and therapeutic footwear[40] during running and gait trials. Differences in subjects, the construction of the FO, and functional activities investigated in each of the papers may account for the conflicting results.

Kulig *et al.*[24] reported increased selective activity of tibialis posterior with an FO in contrast to a barefoot baseline condition, during a resisted foot adduction exercise. Although direct comparisons cannot be made between papers, the findings of Kulig *et al.*[24] provide further evidence that various types of FOs significantly affect the extrinsic foot muscles (including tibialis anterior and posterior, medial and lateral gastrocnemius, peroneals and soleus), which have long been associated with control of frontal plane motion of the foot, particularly those acting primarily as inverters.

The peroneal muscles do not control rearfoot pronation but rather act to evert the ankle and control ankle inversion; however, they are grouped within the complex of 'extrinsic foot muscles'. Both Mündermann *et al.*[21] and Murley and Bird[25] reported increased peroneus longus activity with rigid FOs during running and walking, further indicating the effects FOs have on this collective group of foot inverter muscles.

Evidence for the effect of FOs on thigh muscle activity is less conclusive, with several papers reporting a general increase in vastus medialis activity with FOs compared to barefoot or control conditions,[21,22,39,40] although this increase was not statistically significant in all cases. In contrast, Nurse *et al.*[28] and Nawoczenski and Ludewig[23] have reported no significant differences in vastus medialis activity between orthotic and non-orthotic conditions. Similar conflicting results are also cited for vastus lateralis,[21,23,39] biceps femoris[21,23,28,40] and rectus femoris.[28,39] The absence of a significant effect on the EMG profiles of these muscles between orthotic and non-orthotic conditions could suggest that individuals have compensated for a change in lower limb alignment with functional changes in the foot and ankle musculature or other muscles, which were not monitored during these studies.

Importantly, alterations in the performance of functional activities may have been brought about by changes in activation of more than one muscle.

However, there is insufficient information in the papers reviewed to clearly determine how systems of synergist or antagonist muscles in the lower limb are affected. It is therefore unclear whether changes in performance were attributable to changes in isolated muscles or group patterns, so further research should be carried out in this area.

The studies reviewed showed some similarities between the lower limb muscles but the lack of consensus in methodological approaches, FOs, outcome measures, data acquisition and data extraction mean that the studies were not comparable. The differing construction material and design of each FO type may affect muscle activation in different ways. Based on the current results it is proposed that three different mechanisms may exist by which muscle activation patterns may be altered when wearing FOs: motion control, shock absorbency and enhancing sensory afferent input.

Motion control

Semi-rigid FOs may enhance the coupling effect between the leg and rearfoot. For example, an associated decrease in biceps femoris activity may suggest that when semi-rigid extrinsically posted FOs are worn, this lessens the demands placed on the lateral hamstring to control tibial internal rotation.[23] Alternatively, realigning lower limb muscles, in particular tibialis anterior, into a mechanically advantageous position in preparation for re-supination of the foot prior to terminal stance, could elicit greater muscle activation through an earlier change from eccentric to concentric activity. Enhanced tibialis anterior activity was also proposed to have arisen from the FOs' semi-rigid material,[23] which could potentially compromise subtalar joint stability. This would increase the demand on the muscle to control rapid forefoot descent following heel contact and counteract instability. However, conflicting evidence from Tomaro and Burdett[26] suggests that, as tibialis anterior and medial gastrocnemius act primarily on sagittal plane movement at the talocrural joint, the non-significant changes reported in average EMG activity between rigid FO and non-FO conditions reflect the role of FOs to control subtalar joint pronation with the greatest effect being evident in the frontal plane of the calcaneus. The findings of Kulig *et al.*[24] and Vanicek *et al.*[27] demonstrate enhanced activity in a specifically-targeted muscle, accompanied by reduced supplementary muscle activity i.e. activity in additional muscles required in order to complete a task, and reduced muscle fatigue in the presence of a soft/semi-rigid FO. This suggests that joint stability is a major strategy for reducing muscle activity. Therefore, an FO that results in minimal muscle activity to maintain a stable movement or position could minimise muscle fatigue.

Murley and Bird[25] postulated that rigid FOs may assist the inherent 'windlass mechanism' of the foot. This mechanism is one by which the foot can raise the longitudinal arch, supinate the rearfoot and become a more stable structure. During gait, as the hallux dorsiflexes, the plantar fascia is pulled around the first metatarsal head (the windlass). The winding of the plantar fascia consequently shortens the distance between the calcaneus and metatarsophalangeal joints, causing the metatarsal to plantarflex and the medial longitudinal arch to be raised.[43] The cumulative effects of the flexor digitorum longus, flexor hallucis longus, peroneus longus and Achilles tendon permit the degree of supination required to enhance the windlass mechanism.[44] Weakness in gluteus medius, gluteus minimus, tensor fascia latae, or quadriceps muscles can contribute to abnormalities with the plantar fascia and windlass mechanism. Proximal muscle weakness will lead to inefficient shock absorption and pronation control, as the ability of the muscles to assist in lower limb load response is decreased, causing greater shock transmission to the supporting foot structures.[45]

The role of therapeutic footwear is open to debate. In agreement with Nawoczenski and Ludewig,[23] the findings of Romkes *et al.*[39] suggest that, akin to semi-rigid FOs, soft-soled shoes can influence lower limb kinetics and co-contraction muscle activation patterns. The soft heel, rocker bottom, and raised position of the ankle created by these shoes form an unstable base of support. This may elicit changes in the intensity and/or timing of muscle activity to maintain agonist-antagonistic contractions and enhance foot-ankle joint stability on impact.

Shock absorbency

Changes in the lower limb EMG activity with FO interventions may indicate that lower-extremity muscles have a role in dampening soft tissue vibrations.[21] In addition to the rigid posted FO used by Mündermann *et al.*,[21] this response has been elicited through the elastic and viscous-elastic midsole material comprising shoe heels.[46] Higher vertical ground reaction forces and loading rates after heel strike have also been observed with posted and moulded plus posted FOs,[13] demonstrating the shock-absorbing and 'muscle-tuning' assets of semi-rigid FOs.

Soft tissue masses will vibrate in a dampened manner in response to oscillation caused by direct impact force, for example as the foot collides with the ground during locomotion.[13] However, vibration characteristics of soft tissue can be altered via muscle activity. Natural frequencies of vibrational muscle activity in the triceps surae, quadriceps and tibialis anterior are cited to be in the range of 10 Hz in a state of relaxation, to 50 Hz when fully active.[13] Within 50 ms after heel-strike, shock waves produced on impact travel through both soft tissues and skeletal components. If the natural frequency of the muscle is close to the frequency of the input force, the tissues may resonate.[47] Peak impact forces during heel-toe running can have an input force within the range of 10–20 Hz.[48] Therefore in response to mechanical stimuli such as impact shock, it would be expected that soft tissue resonance would occur at this phase in the gait cycle. Continual exposure to superimposed vibrations is detrimental to soft tissue: it reduces motor unit firing rates, contraction force during intermittent and sustained maximal voluntary contractions and force fluctuations during sub-maximal isometric contractions.[48–51] Subsequently, muscles of the lower limb may have an inherent pre-programmed strategy to minimise soft tissue resonance. Such muscle activity in the lower extremities, responding to impact shock at heel strike during walking or running, is referred to as 'muscle-tuning'.[40]

Enhancing sensory afferent input

Hertel *et al.*[22] postulate that FOs may alter plantar afferentation patterns which thereafter influence muscle activation during functional activities. However, speculation arises concerning the mechanism by which FOs initiate these changes. It has been suggested that postural responses to tactile stimulation from multiple plantar regions of the foot may be co-processed and integrated by the central nervous system (CNS).[52] Kavounoudias *et al.*[52] demonstrated that concurrently stimulating two areas of the plantar surface of the foot resulted in a postural response, which was the sum of the responses when each plantar region was stimulated separately. Therefore, processing multiple inputs of tactile stimuli may enable the CNS to obtain a spatial distribution cue of the pressures applied to the feet, which can then be transformed into a postural response of a specific direction and amplitude.[52]

Other studies suggest that textured soft FOs[28] and semi-rigid and rigid FOs[21] may change the rate of discharge from mechanoreceptors, or the spatio-temporal firing patterns of populations of sensory afferents. It has been demonstrated that during low intensity ambulation, textured inserts have elicited greater change in low- relative to high-frequency EMG signals, which correspsond to slow motor unit pools.[28] In contrast, greater relative change in terms of EMG intensity has been observed in the high-frequency EMG signal (which corresponds to fast motor unit pools) with semi-rigid and rigid FOs during intense ballistic activity.[21] This may suggest that the effect of FOs on general lower limb muscle activity and, more discretely, the effect on slow or fast motor unit pools, is specific to the intensity of the task. Furthermore, body movement in any direction and of any amplitude causes lengthening of specific muscles. This is coupled with pressure increases in one or more zones of the sole,[50] and may cause changes in muscle activity under conditions of upright stance. However, it is unknown whether the decreased muscle activity observed by Nawoczenski and Ludewig[23] and Nurse *et al.*[28] during dynamic tasks indicates a more efficient movement pattern or a decrease in performance. It is therefore speculative whether increased sensory feedback would be desirable in all subject populations, i.e. elite athletes, young people and older healthy adults. Bird *et al.*[30] demonstrated that rigid foot wedging significantly alters the onset of muscle activity in the spine and pelvic regions during ambulation. However, one cannot discern whether such changes are indirectly related to biomechanical alteration of the lower kinetic chain, enhanced afferent feedback to the plantar aspect of the feet, or whether foot wedges facilitate the shock absorbing ability of the body to counteract potential spinal damage.

Conclusion

FOs have been shown to both increase and decrease lower limb muscle activation within specific muscles, in terms of EMG intensity, onset of muscle activity and co-contraction patterns. However, there is limited consensus regarding the direction of change of muscle activity. From this review, it is evident that a number of theories exist concerning the mechanisms by which FOs can alter muscle activity in the lower limbs, including motion control, shock absorbency, and/or enhancing afferent sensory feedback at the plantar surface of the foot. It remains unknown whether a specific property within the design of FOs has greatest influence on muscle activation, or whether their geometrical contours, textured surface or degree of compression make any significant difference. The majority of current research investigating the effect of FOs on lower limb EMG muscle activity largely focuses on healthy, young, recreationally active adults. There is limited information regarding the effect of FOs on lower limb EMG muscle activation in the older adult. Further studies including trials that include a representative sample of the general population are needed to determine the clinical utility of this approach in understanding the role of foot orthoses and footwear in lower muscle activity.

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Participant No:

UNIVERSITY OF TEESSIDE
SCHOOL OF HEALTH & SOCIAL CARE

CONSENT FORM

Title of Study: Do textured surfaces affect postural stability and lower limb muscle activity in healthy adults?

Name of Researchers: Anna Hatton, Dr John Dixon

Please initial all boxes

1. I confirm that I have read and understand the information sheet dated _________ for the above study and have had the opportunity to ask questions. ☐

2. I understand that my participation is voluntary and that I am free to withdraw at any time. Any information collected to that point will be destroyed. ☐

3. I understand that I am at minimal risk participating in this study, but it is possible that I may have a minor allergic reaction to the hypoallergenic tape. ☐

4. I understand that all information from the study will be stored securely, and locked away in a secured cabinet. ☐

5. I understand that all my information will be confidential, my anonymity will be maintained at all times and I will not be identified in any report or publication. ☐

6. I understand that my data will be stored in a fireproof locked cabinet based at the Centre for Rehabilitation Sciences, then destroyed after the study has been completed. ☐

7. I agree to take part in this study ☐

______________________	__________	________________________
Name of Participant	Date	Signature
____________________	_________	_____________________
Name of Researcher	Date	Signature

Do Textured Surfaces Affect Balance and Lower Limb Muscle Activity in Healthy Adults?

Why is this Research Important?

This study is being carried out as part of a research thesis for an MPhil/PhD postgraduate degree. The purpose of this research is to assess the effects of a variety of textured surfaces on balance control and leg muscle activity in healthy adults. This information is important as it may help to identify future ways in which falls prevention could be addressed.

What would be involved?

- You would be required to attend the Physiology Lab (C2.15) for approximately 45 to 60 minutes on two occasions, one week apart.
- You will undertake measures of quiet standing balance on the Kistler Force Platform with eyes open and eyes closed on three different textured surfaces.
- You will then perform a sit to stand movement from a chair again on the Kistler Force Platform on all three surfaces. EMG recordings of 8 lower limb muscles will be made throughout all tests.

What's in it for you?

- An opportunity to familiarise yourself with some pieces of kit available in the physiology lab.
- Involvement in a project, which may generate ideas for your own dissertation.

Anyone Interested? All BSc / MSc Physiotherapy Students Welcome!

Contact: Anna Hatton (a.hatton@tees.ac.uk) Tel: 01642 342934 / 07952 158930

UNIVERSITY OF
TEESSIDE
SCHOOL OF HEALTH
& SOCIAL CARE

PHASE 1 PARTICIPANT INFORMATION LEAFLET

Title of Study: Do textured surfaces affect balance control and lower limb EMG muscle activity in healthy young adults?

Invitation:

You are being invited to take part in a research project. Before you decide to participate it is important that you understand why the research is being undertaken and what your participation in the study will involve. Please read the following information carefully. You may wish to discuss the matter with others before making your decision. The researcher will be happy to answer any questions you may have about the study, before you decide whether to participate or not.

What is the purpose of the study?

The purpose of the study is to assess the effects of a variety of textured surfaces on balance control and leg muscle activity in healthy adults. This information is important as it may help to identify future ways in which falls prevention could be addressed.

Why have I been chosen?

You have been chosen to participate in this study as you have voluntarily responded to the recruitment poster requesting healthy adults. In accordance with the exclusion criteria listed on the participant information leaflet, you **do not**: have Parkinson's Disease, Multiple Sclerosis or have had a Stroke; type II diabetes, a lack of sensation on the soles of your feet, an inner ear disorder, vertigo or dizziness, are unable to walk 10 metres unassisted and/or stand up independently from a chair without using your hands.

Do I have to take part?

It is your decision whether to participate or not in the study. If you do decide to participate you will be asked to sign a consent form. You will be free to withdraw from the study at any time without giving a reason.

What will happen to me if I take part?

After consent is given, you will be asked to attend the Physiology Laboratory at the University of Teesside for approximately 60 minutes on two occasions, one week apart. You will be asked to remove all footwear, socks/tights and to either wear loose fitting shorts, a skirt, or roll up your trouser leg for this duration. Small areas of skin on your dominant leg will need to be cleaned using isopropyl alcohol swabs. For optimal electrode-skin contact, these areas on your thigh and lower limb may also need to be shaved using an electric razor, before electrodes can be attached. These procedures will be pain-free and will not cause any discomfort. Recording surface electrodes will then be placed over the muscles of interest at specified sites whilst you are standing. Gel will be applied to the electrodes, before they are fixed to your skin with strips of hypoallergenic micropore tape. You may feel a cold sensation as the gel makes contact with your skin.

What do I have to do?

Initially, you will be asked to perform some simple movements of your leg against a moderate resistance, provided manually by the researcher. This ensures the electrodes are positioned correctly over the muscles. You will be instructed to stand barefoot for 30 seconds with your eyes open and then closed. The researcher will announce when testing has begun and when it is over.

What are the possible disadvantages and risks of taking part

There are no serious side effects or risks anticipated for participants taking part in the study. It is possible that you could have a minor allergic reaction to the hypoallergenic tape used on your leg. If you know of any allergies that you could have to the tape, please bring this to the attention of the researcher. The area where the electrodes are to be placed may need to be shaved. For safety purposes the researcher, who is a qualified physiotherapist, will remain at your side throughout the testing procedures. Any problems should be reported to the Director of Studies, Dr. John Dixon on (01642) 384125

What if something goes wrong?

All parties in the study are covered by the university’s indemnity insurance.

What about confidentiality?

All information collected about you during the study will be confidential and will be securely stored and locked away in a fireproof cabinet based at the Centre for Rehabilitation Sciences. Your anonymity will be maintained at all times.

What will happen to the results of the study?

The results of the study will be written up and submitted towards an MPhil/PhD postgraduate degree at the University of Teesside. You will not be identified in any report or publication of the study results. All data will be destroyed at the end of the study.

Who has organised and reviewed the study?

This study is being carried out as part of a research thesis for an MPhil/PhD postgraduate degree at the University of Teesside. Ethical approval has been sought from the School of Health and Social Care Research Ethics Committee, University of Teesside.

Thank you for taking time to read this information leaflet.

For further information please contact: Anna Hatton or Dr John Dixon,
Centre for Rehabilitation Sciences,
School of Health & Social Care,
University of Teesside,
Middlesbrough, TS1 3BA.
Tel: (01642) 342934/384125

Participant No:

UNIVERSITY OF TEESSIDE
SCHOOL OF HEALTH & SOCIAL CARE

PHASE 1 PARTICIPANT DETAILS

1. **Date of Birth**:
2. **Gender** *(delete as applicable):* **MALE / FEMALE**
3. **Weight** *(kg)*:
4. **Height** *(cm)*:
   Knee Height *(cm):*
   Eye Height *(cm):*

5. **In your own words, describe your *current* general health and fitness. Include details regarding any surgery you have undergone recently** *(within the last 12 months)*:

___

___

6. **Exclusion Criteria** *(please tick the box below if any one or more of the following statements apply to you):*

- You have Parkinson's disease or multiple sclerosis.
- You have had a stroke.
- You have a lack of sensation on the soles of your feet.
- You have type II diabetes.
- You have an inner ear disorder.
- You have vertigo or dizziness.
- You require assistance to walk 10 metres/11 yards.
- You are unable to stand up and/or sit down independently from a chair, without using your hands.

Please tick here if any one or more of the statements above apply to you: ☐

Participant No:

UNIVERSITY OF TEESSIDE
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7. **Visual Problems** *(delete as applicable):*

- You currently wear glasses and/or contact lenses: **YES / NO**

8. **Foot Posture Index:**

	Factor	**Plane**	**Score**	
			Left -2 to +2	**Right -2 to +2**
Rearfoot	Talor head palpation	Transverse		
	Curves above and below lateral malleolus	Frontal/Transverse		
	Inversion/eversion of the calcaneous	Frontal		
Forefoot	Prominence in the region of the TNJ	Transverse		
	Congruence of the medial longitudinal arch	Sagittal		
	Abduction/adduction forefoot on rearfoot	Transverse		
		Total		
		Foot Posture		

Participant No:

UNIVERSITY OF TEESSIDE
SCHOOL OF HEALTH & SOCIAL CARE

9. **Monofilament Testing** *(Maximum x3 applications per site per foot):*

Location	Response (Right Foot)	Response (Left Foot)
Between the First and Second Metatarsals (Dorsal Surface)		
Great Toe (Plantar Surface)		
First Metatarsal Head (Plantar Surface)		
Fifth Metatarsal Head (Plantar Surface)		
Heel		
	Insensate / Sensate	**Insensate / Sensate**

APPENDIX 6: Descriptive statistics for postural sway variables during quiet standing with eyes open over 30 seconds in healthy young adults (n=24)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	4.4 (1.4)	3.9	1.3	4.1 (0.9)	4.0	0.9	4.6 (0.8)	4.3	1.3
AP range (mm)	24.5 (5.7)	22.7	5.0	23.1 (3.6)	22.8	3.5	24.8 (3.0)	25.5	5.8
ML SD (mm)	3.0 (0.7)	3.3	1.1	3.1 (0.7)	3.3	1.1	3.0 (0.6)	3.2	0.9
ML range (mm)	18.2 (3.4)	17.7	6.5	18.2 (3.3)	18.6	4.6	18.3 (2.9)	18.3	4.5
CoP velocity (mm s^{-1})	9.2 (3.2)	8.4	2.6	10.4 (5.5)	8.4	4.1	10.8 (5.0)	8.1	7.1

APPENDIX 7: Descriptive statistics for postural sway variables during quiet standing with eyes closed over 30 seconds in healthy young adults (n=24)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	4.8 (1.2)	4.4	1.2	4.7 (1.0)	4.6	1.3	4.8 (1.1)	4.6	1.3
AP range (mm)	27.1 (6.0)	26.4	4.1	27.7 (5.3)	27.0	9.4	27.8 (4.7)	28.1	8.1
ML SD (mm)	3.1 (0.6)	3.1	0.9	3.1 (0.8)	3.2	1.2	3.2 (0.6)	3.2	1.0
ML range (mm)	18.8 (3.1)	19.2	4.3	19.5 (4.3)	18.4	6.6	19.4 (3.1)	20.0	4.5
CoP velocity (mm s^{-1})	13.2 (7.4)	10.6	8.4	11.5 (4.4)	10.5	3.3	12.9 (7.4)	10.8	5.4

APPENDIX 8: Descriptive statistics for postural sway variables during quiet standing with eyes open over the first 10 seconds in healthy young adults (n=24)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	3.4 (0.7)	3.3	0.8	3.6 (0.9)	3.6	1.2	3.6 (0.7)	3.6	1.0
AP range (mm)	17.9 (3.4)	17.3	4.2	19.0 (3.8)	18.8	5.4	18.9 (3.1)	18.8	4.0
ML SD (mm)	3.0 (0.7)	3.2	1.2	2.9 (0.6)	3.1	1.0	2.9 (0.6)	3.0	1.1
ML range (mm)	16.2 (3.1)	15.8	5.1	15.8 (2.8)	15.3	5.0	15.8 (2.9)	15.5	4.5
CoP velocity (mm s^{-1})	10.8 (3.6)	9.8	1.9	12.4 (5.7)	10.8	4.4	12.1 (5.1)	10.1	5.8

APPENDIX 9: Descriptive statistics for postural sway variables during quiet standing with eyes closed over the first 10 seconds in healthy young adults (n=24)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	4.3 (1.2)	4.0	1.1	4.2 (0.8)	4.1	1.2	4.2 (0.8)	4.1	1.0
AP range (mm)	22.2 (5.4)	21.8	5.0	22.3 (3.6)	22.1	6.0	22.1 (3.3)	21.8	5.6
ML SD (mm)	3.1 (0.6)	3.2	0.7	3.1 (0.8)	3.0	1.1	3.0 (0.6)	3.1	1.0
ML range (mm)	16.9 (3.3)	16.8	3.6	17.3 (3.6)	17.1	4.2	16.7 (3.1)	16.7	5.0
CoP velocity (mm s^{-1})	16.4 (7.8)	13.4	7.0	14.5 (4.2)	12.7	5.7	15.4 (7.6)	13.2	6.8

APPENDIX 10: Descriptive statistics for postural sway variables during quiet standing with eyes open over the latter 20 seconds in healthy young adults (n=24)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	3.9 (1.1)	3.6	1.0	3.6 (0.7)	3.4	0.8	3.9 (0.7)	3.9	1.0
AP range (mm)	20.5 (5.0)	19.1	4.6	19.2 (2.7)	19.0	3.9	20.6 (3.2)	20.4	4.6
ML SD (mm)	2.7 (0.7)	2.9	1.1	2.7 (0.6)	2.8	1.0	2.8 (0.6)	2.8	1.0
ML range (mm)	14.9 (3.2)	15.6	5.9	14.7 (2.8)	15.0	4.8	15.4 (2.8)	15.2	4.5
CoP velocity (mm s^{-1})	8.4 (3.2)	7.8	2.3	9.4 (5.6)	7.3	4.4	10.2 (5.3)	7.9	6.7

APPENDIX 11: Descriptive statistics for postural sway variables during quiet standing with eyes closed over the latter 20 seconds in healthy young adults (n=24)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	4.1 (0.9)	3.9	1.0	4.1 (1.2)	3.9	1.5	4.2 (1.0)	4.0	0.9
AP range (mm)	21.5 (4.4)	21.4	5.2	22.1 (5.8)	22.1	6.4	22.9 (4.6)	22.1	5.6
ML SD (mm)	2.7 (0.6)	2.8	0.7	2.8 (0.9)	2.9	1.5	2.8 (0.7)	3.1	1.0
ML range (mm)	15.4 (2.8)	15.6	3.0	15.6 (4.4)	15.5	5.6	16.1 (3.5)	16.8	5.2
CoP velocity (mm s^{-1})	11.6 (7.4)	9.4	6.8	10.1 (4.7)	9.3	2.5	11.6 (7.4)	9.6	4.8

APPENDIX 12: Descriptive statistics for lower limb EMG activity (μV) during quiet standing with eyes open over 30 seconds in healthy young adults (n=24)

	Control			**Texture 1**			**Texture 2**		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
Vastus Medialis	12.3 (6.1)	10.2	1.9	12.4 (6.0)	10.2	1.4	12.8 (6.8)	10.3	1.4
Rectus Femoris	3.4 (2.3)	2.4	2.3	3.4 (2.2)	2.3	1.7	3.6 (2.5)	2.7	2.2
Vastus Lateralis	6.4 (9.0)	2.4	4.6	6.7 (8.7)	2.4	6.6	7.6 (10.8)	2.4	7.4
Biceps Femoris	3.6 (1.8)	3.0	2.4	4.2 (3.3)	2.9	3.3	4.5 (3.7)	3.0	3.2
Tibialis Anterior	3.6 (3.1)	2.7	0.8	2.9 (0.8)	2.7	0.6	3.3 (2.8)	2.7	0.8
Peroneus Longus	11.0 (5.5)	10.1	4.9	10.9 (5.4)	9.4	5.2	10.6 (5.6)	8.8	4.6
Medial Gastrocnemius	10.0 (9.2)	7.3	6.8	9.7 (7.0)	7.7	6.5	10.2 (8.9)	8.0	6.3
Soleus	17.8 (9.8)	17.2	14.4	17.5 (8.7)	16.8	11.4	16.9 (8.6)	16.0	10.8

APPENDIX 13: Descriptive statistics for lower limb EMG activity (µV) during quiet standing with eyes closed over 30 seconds in healthy young adults (n=24)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
Vastus Medialis	13.3 (7.0)	10.5	5.7	12.8 (7.0)	10.6	1.3	13.3 (6.9)	11.5	3.4
Rectus Femoris	4.1 (3.0)	2.9	3.3	3.6 (2.7)	2.7	2.1	4.0 (2.6)	3.4	2.6
Vastus Lateralis	7.9 (11.4)	2.6	9.4	7.8 (11.7)	2.9	9.1	8.2 (11.4)	4.9	7.8
Biceps Femoris	4.5 (3.5)	3.2	3.7	5.4 (6.1)	3.3	2.4	4.0 (2.7)	3.1	2.7
Tibialis Anterior	4.0 (3.5)	2.9	1.1	3.4 (2.1)	2.9	1.0	3.8 (3.7)	2.9	1.1
Peroneus Longus	11.1 (5.4)	9.6	7.3	11.5 (5.4)	9.9	7.2	11.5 (5.1)	9.4	5.8
Medial Gastrocnemius	11.6 (8.6)	9.5	9.0	11.7 (8.4)	10.4	9.2	10.5 (7.1)	10.3	6.8
Soleus	17.6 (8.7)	16.5	7.0	18.4 (8.9)	17.1	8.8	18.1 (8.7)	16.4	10.9

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The effect of textured surfaces on postural stability and lower limb muscle activity

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Abstract

Textured insoles may enhance sensory input on the plantar surfaces of the feet, thereby influencing neuromuscular function. The aim of this study was to investigate whether textured surfaces alter postural stability and lower limb muscle activity during quiet bipedal standing balance with eyes open. Anterior–posterior (AP) and mediolateral (ML) sway variables and the intensity of electromyographic (EMG) activity in eight dominant lower limb muscles were collected synchronously over 30 s in 24 young adults under three randomised conditions: control surface (C), texture 1 (T1) and texture 2 (T2). Repeated measures ANOVA showed that the textured surfaces did not significantly affect AP or ML postural sway in comparison to the control condition ($p > 0.05$). Neither did the textured surfaces significantly alter EMG activity in the lower limbs ($p > 0.05$). Under the specific conditions of this study, texture did not affect either postural sway or lower limb muscle activity in static bipedal standing. The results of this study point to three areas of further work including the effect of textured surfaces on postural stability and lower limb muscle activity: (i) in young healthy adults under more vigorous dynamic balance tests, (ii) post-fatigue, and (iii) in older adults presenting age-related deterioration.

Keywords: Texture; Sensory input; Postural stability; EMG activity

1. Introduction

Recent research indicates that footwear interventions, including foot orthoses and insoles, can influence both static and dynamic functional ability (Nawoczenski and Ludewig, 1999; Bird et al., 2003; Vanicek et al., 2004; Hertel et al., 2005; Kulig et al., 2005; Nurse et al., 2005; Murley and Bird, 2006; Romkes et al., 2006).

For example, vibrating insoles have been shown to affect short-term postural sway variables in young and older adults (Priplata et al., 2003) and semi-rigid, rigid and soft foot orthoses have been shown to alter lower limb electromyographic (EMG) activity in terms of intensity (Tomaro and Burdett, 1993; Nawoczenski and Ludewig, 1999; Hertel et al., 2005; Kulig et al., 2005; Mündermann et al., 2006; Murley and Bird, 2006), onset of activity (Bird et al., 2003) and fatigue (Vanicek et al., 2004), during dynamic functional activities.

As well as changes in biomechanical alignment, motion control and facilitation of shock absorbance, enhanced sensory input to the plantar aspect of the feet has been discussed as a factor in footwear interventions (Hatton et al., 2008). It has been proposed that the presence or texture of footwear interventions may enhance sensory input influencing lower limb muscle activity during dynamic function (Hertel et al., 2005; Nurse et al., 2005) with the proposed mechanisms being alterations in the rate of discharge from mechanoreceptors, or spatio-temporal firing patterns of populations of sensory afferents in the plantar surface of the feet (Watanabe and Okubo, 1981; Waddington and Adams, 2003; Nurse et al., 2005; Wilson et al., 2008). Nurse et al., 2005 investigated the effect of textured insoles comprising semi-circular mounds with centre-to-centre

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distances of 8 mm, on gait patterns and reported significant reductions in tibialis anterior and soleus EMG activity. Waddington and Adams (2000, 2003) investigated the effect of textured versus smooth insoles on movement discrimination of ankle inversion. When substituted for smooth insoles, textured insoles restored movement discrimination back to barefoot levels. Watanabe and Okubo (1981) found that postural sway was significantly reduced when standing on a textured surface in comparison to a smooth surface. However, more recently, textured insoles with dimpled and grid surfaces have shown to have no significant effect on postural sway variables in healthy older women over a four-week period (Wilson et al., 2008). The research base on the effect of textured insoles is, therefore, limited. Also, to our knowledge, no studies have evaluated the effect of texture on static postural stability and lower limb muscle activity simultaneously.

The aims of this study were to determine whether textured surfaces alter postural stability and EMG activity during quiet standing balance in young, asymptomatic adults.

It was hypothesised that among the conditions of control (flat surface) and two textured surfaces there would be:

1. differences in the magnitude of postural sway variables, and
2. differences in the amplitude of lower limb EMG activity in eight lower limb muscles.

2. Methods

2.1. Participants

A convenience sample of 24 asymptomatic adults (17 female) took part in the study. Basic details of the sample are as follows: mean age 27.5 (standard deviation 7.9) years; mean height 167 (standard deviation 6) cm; mean weight 67.1 (standard deviation 11.8) kg; mean body mass index 24.1 (standard deviation 4.1). Inclusion criteria were young adults from the university and local community. Exclusion criteria were a self-reported history of neuromuscular disease, stroke, peripheral sensory neuropathy, type II diabetes, inner ear disorders, vertigo, dizziness, musculoskeletal injury or pain in the lower limbs or back, surgery of any kind in the 12 months prior to testing, inability to walk 10 m unassisted, and/or inability to stand up and sit down from a chair without using their hands. Ethical approval was granted by the School of Health and Social Care Research Governance and Ethics Committee at the University of Teesside. Written informed consent was obtained from all participants meeting the inclusion criteria. Participants were permitted to withdraw from the study at any point in the proceedings.

2.2. Design

The study used a within-subject experimental design with participants taking part in testing in each of three conditions – control (C), texture 1 (T1) and texture 2 (T2). The dependent variables were anterior–posterior (AP) postural sway range, AP postural sway standard deviation, mediolateral (ML) postural sway range, ML postural sway standard deviation and average integrated EMG for each of eight lower limb muscles.

2.3. Textured surfaces

Three different textured surfaces were provided by the orthotic supplies company Algeos (Algeos UK Ltd., Liverpool, United Kingdom). T1 (Evalite Pyramid EVA, 3 mm thickness, Shore value A50) and T2 (nora®Lunsasoft Mini Non Slip, 3 mm thickness, Shore value A50) were selected from the range of EVA Soling materials. T1 had small pyramidal peaks whilst T2 displayed convex circular patterning. The control condition used a completely flat surface texture (Medium Density EVA, 3 mm thickness, Shore value A50).

These surfaces were cut manually from commercially available sheets to smaller dimensions of 428 mm × 630 mm × 3 mm to cover the top surface of a Kistler force plate. Three textured surfaces were chosen for evaluation on the basis that T1, comprising small, pointed, pyramidal indentations may depress a smaller area than T2, which comprised slightly larger convex circular patterning. The smooth, flat C texture had no indentations on its surface. It was postulated that the variety of indentation and therefore stimulus provided by the three textures may initiate differing effects on postural sway and muscle activity. All the textures selected have been used in previous research in our laboratory to investigate their effect on postural stability in middle-aged females (Wilson et al., 2008).

2.4. Postural stability

Data for static postural sway was obtained from a Kistler force platform (Model 9286AA, Kistler, Alton, UK), and recorded at a sampling rate of 1000 Hz.

2.5. Electromyography

All EMG recordings were sampled at a rate of 1000 Hz using a 16-channel Biopac system (Model MP 150, BIOPAC Systems Inc., Santa Barbara, CA, USA), using bipolar active surface EMG recording electrodes (Type TSD 150B, 11.4 mm diameter, electrode spacing 20 mm), with 3 dB 12–500 Hz bandpass and ×350 built in amplification.

EMG recordings were collected from eight muscles of the dominant lower limb of the participants, determined by asking participants which leg they would kick a football with. All EMG recording electrodes were then attached to that leg (Rose et al., 2002; Kulig et al., 2005). The eight muscles were: vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA), peroneus longus (PL), medial gastrocnemius (MG) and soleus (SOL). The skin over the muscles of interest was shaved using a disposable razor, if necessary, and cleaned with surgical spirit before electrode placement. Hypoallergenic conducting gel was applied to the electrodes to ensure optimal electrical contact with the skin surface. A pre-gelled ground reference electrode (Blue Sensor®) was placed at the tibial tuberosity on the non-dominant lower limb. Standardisation of electrode placement followed the recommendations of the European project 'surface EMG for non-invasive assessment of muscles' (Hermens et al., 2000; Figs. 1 and 2). Accuracy of electrode placement and absence of cross-talk was confirmed by conducting

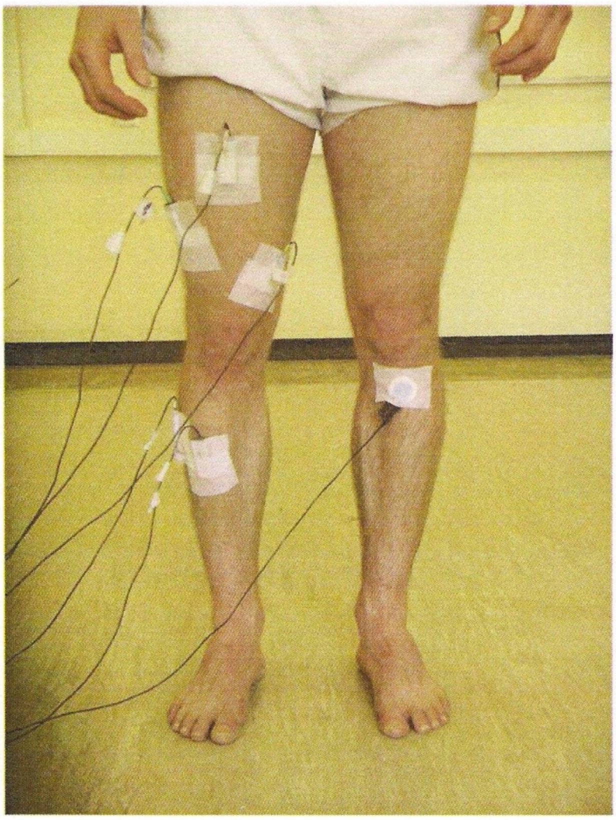

Fig. 1. EMG electrode placement – anterior view.

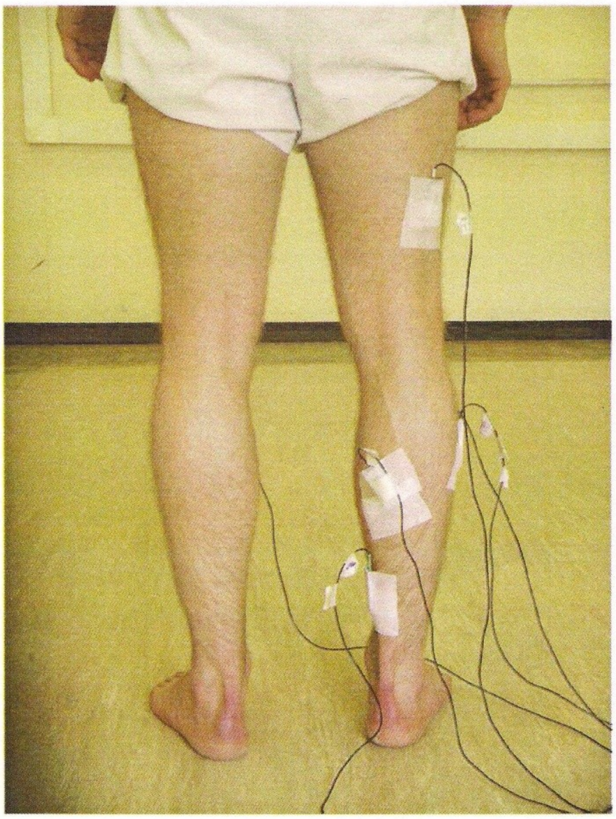

Fig. 2. EMG electrode placement – posterior view.

pain-free lower limb movement against manual resistance provided by the tester (ALH) whilst observing raw EMG waveforms.

The temperature of the laboratory remained constant at 22 °C throughout all testing procedures, to prevent changes in lower limb EMG activity and force with variations in room temperature (Bell, 1993).

2.6. Procedures

All participants were tested in barefoot, bipedal quiet standing with eyes open. Each test repetition lasted for 30 s duration. There were three trials for each condition – a total of nine trials per participant. The participant completed all three trials of one condition before testing took place under another condition. The sequence of test condition followed by the participant was randomised using number cards selected by that participant. Participants also conducted practice trials of quiet standing balance tests to ensure familiarity with the testing procedures, prior to data collection.

Foot posture was assessed for all participants whilst barefoot using the foot posture index (Redmond et al., 2006).

To ensure that the foot position was standardised for each trial the following procedure was adapted from that used by Buckley et al. (2005). An individual template was drawn for each participant according to their preferred barefoot quiet standing position. This template was laid over the textured surface before each trial then removed once the feet had been correctly positioned (Buckley et al., 2005). Participants sat on the end of a height-adjustable plinth and then stood up keeping their feet in the position determined by the template. The template was re-laid on the textured surface following each trial to reposition the feet (Fig. 3).

During testing, participants were instructed to stand with their arms by their side, looking straight forwards and to focus on the middle of a visual target directly ahead, in order to control head positioning and prevent vestibular disruption (Brandt et al., 1981). The visual target was positioned 3 m from the centre of the force plate at eye height.

When the tester (ALH) considered that the participant was ready – trunk was perceived to be vertical and the lower limbs were fully extended – she pressed a trigger that began data collection simultaneously from the EMG system and the Kistler force plate.

Following each test, participants were asked to sit down and take their feet off the textured surfaces and force plate. All participants sat quietly in their normal comfortable position, with their feet resting on either side of the force plate, on the laboratory floor, for two minutes between all tests. This rest period served to prevent habituation to the sensory stimulus of the textured indentations, whilst allowing the next test condition to be prepared and the force plate re-calibrated, prior to data collection.

To allow an evaluation of the between-session repeatability of the data all participants were re-measured at a subsequent session approximately one week later under an identical protocol. All data collection and analysis was carried out by the same investigator.

Fig. 3. Template for foot positioning.

2.7. Data extraction and analysis

AP and ML range and standard deviation (mm) were calculated automatically by the force plate system.

EMG data was processed using dedicated AcqKnowledge software (Version 3.7.3, BIOPAC Systems Inc., Santa Barbara, CA, USA). EMG amplitude (μV) was recorded from all eight dominant lower limb muscles of interest. Using the AcqKnowledge 3.7.3 software, a 20 Hz high pass filter was firstly applied to the raw data to remove any possible movement artefacts and then the raw EMG was processed using a root mean square moving window of 30 ms. For each muscle, the average rectified value (ARV) (Merletti, 1999) was calculated by dividing the EMG integral by the time interval. The data were visually checked to ensure artefacts did not affect the results. The EMG ARV values for each muscle were then averaged (mean) over the three trials for each condition. In this study, EMG normalisation was not required because the participants acted as their own control and all procedures were performed in the same session, without the electrode positions being altered (Soderberg and Knutson, 1995).

2.8. Statistical analysis

Data were analysed using the Statistical Package for Social Sciences (SPSS, Chicago, IL, USA) version 13.0. Between-session repeatability for postural sway and EMG variables occurred within one week of the first session. Data was assessed for the baseline control condition only, by calculating the typical error (Batterham and George, 2003), also known as the standard error of measurement. For each of the variables, a repeated measures analysis of variance (ANOVA) was carried out to determine statistically significant differences among the three conditions with an alpha level set at 0.05. Where the assumption of sphericity was violated, a Greenhouse–Geisser correction was applied. Differences between each pair of conditions were presented as means and 95% confidence intervals (CI).

3. Results

3.1. Between-session repeatability

3.1.1. Postural sway

All postural sway variables showed acceptable between-session repeatability with little variation, as indicated from the measures of typical error (mm): AP standard deviation, 1.1; AP range, 4.9; ML standard deviation, 0.6; ML range, 2.5.

3.1.2. Emg

Again, between-session repeatability for measures of EMG amplitude for all eight lower limb muscles during quiet standing balance on the control surface, showed little variation as indicated by the small typical errors (μV): VM, 0.6; RF, 0.5; VL, 1.9; BF, 2.2; TA, 2.2; PL, 2.8; MG, 4.0; SOL, 3.6.

4. Effect of texture

4.1. Postural sway

From the repeated measures ANOVAs there were no statistically significant differences among the three conditions for any of the postural sway variables: AP SD (F [2,21] = 2.366, $p = 0.105$), AP range (F [2,21] = 1.583, $p = 0.216$), ML SD (F [2,21] = 0.406, $p = 0.669$) and ML range (F [2,21] = 0.021, $p = 0.957$). On inspection of the 95% CI of the differences between each pair of conditions there is a clear asymmetry in the spread of the figures about zero that is in the direction of higher values in T2 compared to T1 for AP range and AP SD. (Indeed for AP standard deviation the 95% CI does not contain zero.) However, any such suggestion is very tentative given the lack of statistical significance shown by the overall ANOVAs. The 95% CIs for the other variables do not support similar speculation for the other variables (Table 1).

4.2. Emg

From the repeated measures ANOVAs there were no statistically significant differences among the three conditions for any of the eight muscle EMG variables (Fig. 4): VM (F [2,21] = 0.727, $p = 0.419$), RF (F [2,21] = 0.930, $p = 0.402$), VL (F [2,21] = 1.042, $p = 0.338$), BF (F [2,21] = 2.967, $p = 0.086$), TA (F [2,21] = 1.314, $p = 0.267$), PL (F [2,21] = 0,772, $p = 0.468$), MG (F [2,21] = 0.187, $p = 0.744$) and SOL (F [2,21] = 1.302,

Table 1
Mean (SD), mean difference (95% CI) of AP range, AP SD, ML range and ML SD during quiet standing balance on three textured surfaces: C, T1 and T2 (mm)

Variable	C mean (SD)	T1 mean (SD)	T2 mean (SD)	C vs. T1 (mean difference and 95% CI)	C vs. T2 (mean difference and 95% CI)	T1 vs. T2 (mean difference and 95% CI)
AP range	24.5 (5.7)	23.1 (3.6)	24.8 (3.0)	−1.4 (−4.3 to 1.5)	0.3 (−2.5 to 3.1)	1.7 (−0.4 to 3.8)
AP SD	4.4 (1.4)	4.1 (0.9)	4.6 (0.8)	−0.3 (−0.9 to 0.3)	0.2 (−0.5 to 0.9)	0.5 (0.01 to 1.0)
ML range	18.2 (3.4)	18.2 (3.3)	18.3 (2.9)	0.03 (−1.6 to 1.6)	0.1 (−1.6 to 1.8)	0.1 (−0.9 to 1.1)
ML SD	3.0 (0.7)	3.1 (0.7)	3.0 (0.6)	0.1 (−0.2 to 0.4)	0.01 (−0.3 to 0.4)	−0.1 (−0.3 to 0.1)

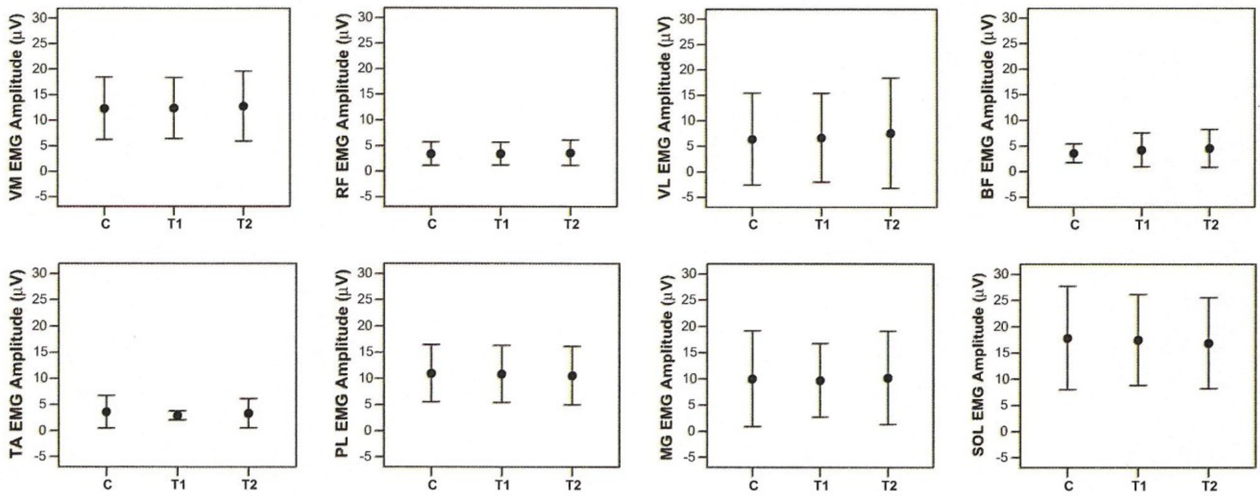

Fig. 4. Mean (SD) EMG amplitude (μV) for vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA), peroneus longus (PL), medial gastrocnemius (MG) and soleus (SOL) during quiet standing balance on three different textured surfaces; control (C), texture 1 (T1) and texture 2 (T2).

Table 2
Mean (SD), mean difference (95% CI) of lower limb EMG amplitude when during quiet standing balance on three textured surfaces: C, T1 and T2 (μV)

Muscle	C mean (SD)	T1 mean (SD)	T2 mean (SD)	C vs. T1 mean difference (95% CI)	C vs. T2 mean difference (95% CI)	T1 vs. T2 mean difference (95% CI)
VM	12.3 (6.1)	12.4 (6.0)	12.8 (6.8)	0.1 (−0.8 to 1.0)	0.5 (−1.0 to 1.9)	0.4 (−0.3 to 1.1)
RF	3.4 (2.3)	3.4 (2.2)	3.6 (2.5)	0.003 (−0.4 to 0.4)	0.2 (−0.2 to 0.5)	0.2 (−0.1 to 0.4)
VL	6.4 (9.0)	6.7 (8.7)	7.6 (10.8)	0.3 (−1.7 to 2.3)	1.2 (−1.6 to 3.9)	0.9 (−0.6 to 2.3)
BF	3.6 (1.8)	4.2 (3.3)	4.5 (3.7)	0.7 (−0.4 to 1.7)	1.0 (−0.4 to 2.3)	0.3 (−0.3 to 0.9)
TA	3.6 (3.1)	2.9 (0.8)	3.3 (2.8)	−0.7 (−2.1 to 0.7)	−0.3 (−0.7 to 0.1)	0.4 (−0.9 to 1.7)
PL	11.0 (5.5)	10.9 (5.4)	10.6 (5.6)	−0.1 (−1.1 to 0.9)	−0.4 (−1.3 to 0.4)	−0.3 (−1.2 to 0.6)
MG	10.0 (9.2)	9.7 (7.0)	10.2 (8.9)	−0.3 (−2.5 to 1.9)	0.2 (−0.9 to 1.2)	0.5 (−1.9 to 2.8)
SOL	17.8 (9.8)	17.5 (8.7)	16.9 (8.6)	−0.4 (−2.3 to 1.5)	−1.0 (−2.7 to 0.7)	−0.6 (−1.5 to 0.4)

$p = 0.276$). Inspection of the 95% CI of the differences between each pair of conditions does not support any contrary speculation (Table 2).

5. Discussion

The purpose of this study was to determine the effect of textured surfaces on static postural stability and lower limb muscle activity in young healthy adults. We found that two textured surfaces, which differed only in their pattern of indentation, did not significantly affect AP or ML postural sway in comparison to the control condition. Neither did the textured surfaces alter the intensity of activity in lower limb muscles.

The results of the current study are also in agreement with those of Wilson et al. (2008) who reported that textured insoles within standardised footwear did not significantly affect postural sway variables over four weeks, in middle-aged females. To isolate the potential effect of textured surfaces from any effects brought about by the construction properties of footwear, including motion control, cushioning and external fastenings, the current study undertook quiet standing balance tests using sheets of texture rather than insoles, on one occasion only, using young adults of mixed gender, whilst barefoot. Together, the results of these studies indicate that textured insoles may not significantly affect postural stability or muscle activity in relatively young people in static balance tests.

Mechanoreceptors respond to mechanical stimuli including indentation and stretching of the skin and can provide information about texture, enabling us to detect roughness, spacing and orientation of textured patterns (Gardner et al., 1991). Small protrusions which indent the skin in localised regions cause a relatively small number of adjacent receptors to fire at high rates. In contrast, gently rounded contours making contact with a larger expanse of skin induce a lower-amplitude firing rate from a larger population of receptors. A smooth flat surface will cause receptors to fire at a constant, low rate (Gardner et al., 1991). The underlying principle of the use of textured surfaces was to enhance sensory input. On the same principle of providing enhanced sensory input to the plantar surfaces of the feet, vibrating insoles have shown to reduce static postural sway in both young and older healthy adults (Priplata et al., 2003). This raises the possibility that any effects are dependent on the nature and/or degree of sensory input.

Had there been changes in postural sway and EMG the design of the study would have allowed us to explore relationships between them. Having reviewed the literature (Hatton et al., 2008) we are not aware of any studies which have investigated the effect of textured insoles or surfaces on lower limb muscle activity in quiet standing balance. However, the textured surfaces had no effect on either postural sway or EMG amplitude. In addition, we did not carry out correlations on the data obtained as the EMG data were not normalised, which meant that between-subject differences in EMG amplitude were not relative, and could just be related to physical differences irrespective of sway.

The current study suggests that textured surfaces do not affect control of bipedal static postural sway or lower limb muscle activity in young, healthy participants. However, under dynamic, as opposed to static, testing conditions texture has been shown to have significant effects. Textured insoles have been shown to decrease overall EMG activity of soleus and tibialis anterior for the entire stance phase during gait trials (Nurse et al., 2005). Previous research also indicates texture insoles can alter ankle movement discrimination (Waddington and Adams, 2003).

The present study investigated two different patterns of indentation (one texture comprising a small pyramidal design and the other a small circular convex pattern) with the aim of providing different quality and/or quantity of sensory input. While, neither texture produced statistically significantly changes in postural sway, the 95% confidence intervals of differences between the two textures for AP range and AP standard deviation tentatively indicate a possible effect of texture type that did not reach statistical significance. Our results indicate that the variety of indentation and thus stimulus provided by the textured surfaces may bring about different effects on postural sway.

5.1. Limitations

It is possible that the balance test in the present study was insufficiently demanding for a young asymptomatic group. Given the findings of Maki et al. (1999, 2008), Nurse et al. (2005) and Mattacola et al. (2007), it may be that the effect of textured surfaces in young healthy participants is apparent only when the balance control system is significantly challenged, for example during dynamic activity, post-fatigue, or when the limits of stability are stretched to their extremes. It is therefore possible that quiet static bipedal standing balance is not sensitive enough to determine whether textured surfaces can alter postural stability and lower limb muscle activity in this specific population or when all sub-systems contributing to balance control, i.e. visual, vestibular, somatosensory, are intact. However, this posture could be considered inherently unstable as a large body mass and high centre of mass must be kept within the boundaries of a small base of support in order to remain upright. There is no consensus of opinion as to which test of balance would be regarded the 'gold standard' to assess static postural control in young healthy adults however, and various studies investigating the contribution of lower limb muscle activity to postural control have adopted the bipedal stance position (Joseph and Nightingale, 1952; Fitzpatrick et al., 1994; Mochizuki et al., 2005; Onambele et al., 2006).

Regarding the EMG findings, it is also possible that in the present study, given the static nature of the balance test and absence of any external perturbation or postural challenge, the muscular response in the lower limbs was minimal in healthy young adults. This may suggest that static balance measures may be more suited to older adults or those with muscular atrophy, decreased muscle strength or nerve conduction speed. We could postulate that age-related degeneration of sensory and neuromuscular control mechanisms could generate greater amounts of postural sway and consequently greater amounts of corrective postural muscle activity and co-activation in the lower limbs.

Given that the current study did not report any change in postural stability whilst standing on different textured surfaces, it is possible that the sway variables used may not be sensitive enough to detect changes in neuromuscular function. However range of CoP displacement has shown to increase following induced ankle muscle fatigue and application of ankle muscle vibration in young asymptomatic adults during bipedal stance (Vuillerme et al., 2002). Previous investigations of postural sway in adults also frequently cite the use of measures of CoP displacement in terms of AP and ML standard deviation (LeClair and Riach, 1996; Caron, 2003; Mochizuki et al., 2005), and AP and ML range (Vuillerme et al., 2002; Priplata et al., 2003; Perry et al., 2007; Wilson et al., 2008). In addition, LeClair and Riach (1996) reported AP and ML standard deviation to increase with increasing testing duration from 10 to 60 s. Hence, it is possible that an increase in test duration may identify greater variability of AP and ML excursion as the target around which participants' sway may change over time.

6. Conclusions

Under the specific conditions of this study, texture did not affect either postural sway or lower limb muscle activity in static bipedal standing. The results of this study point to three areas of further work:

1. Given the direction of the 95% CIs for AP range and standard deviation that suggested a possible difference in effect between the two textures, this part of the study should be replicated with an increase in numbers of participants.
2. Also warranted are investigations into the effect of textured surfaces on postural stability and lower limb muscle activity in young asymptomatic adults under more vigorous dynamic balance tests, post-fatigue of lower limb muscles.
3. Finally, the effects of textured surfaces on postural stability and muscle activity in older adults presenting with age-related deterioration is an area that requires investigation.

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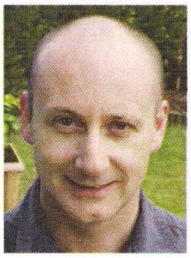
Anna Hatton Anna is a Research Assistant in Rehabilitation Sciences and PhD student at the Centre for Rehabilitation Sciences, University of Teesside. In 2005, she was awarded a BSc (Hons) Physiotherapy from the University of Teesside. Anna's research interests include footwear interventions, falls prevention and neuromuscular function in young and older adults.

John Dixon John has a degree in Physiology and a PhD in Rehabilitation Sciences. His PhD at University of Teesside was an electromyographic analysis of quadriceps muscle function in patients with knee osteoarthritis. His current research includes the prevention and management of musculoskeletal conditions such as osteoarthritis, anterior knee pain and soft tissue injuries of the knee. John is interested in the importance of muscle function in health, disease and rehabilitation, and the relationship between ageing and decline in physical function such as strength and balance.



Denis Martin Denis is the Director of the Centre for Rehabilitation Sciences at the University of Teesside. Denis graduated from the University of Ulster in 1988 with a BSc (Hons) Physiotherapy. He was awarded his DPhil from the University of Ulster in 1993 and received an MSc Applied Statistics from Napier University in 2000. Denis' research interests lie in assessing and managing the impact of pain.

Keith Rome Keith is a Professor in Podiatry at AUT University, New Zealand. He was awarded a Bachelor of Science in Podiatry from the University of Westminster in 1989, a postgraduate diploma in Biomechanics from the University of Strathclyde in 1990, and a Master of Science Degree in Research Methodology for Physical Therapists from Kings College London in 1994. In 2000, Keith was awarded a PhD from the University of Teesside. Keith's research interests include foot and ankle characteristics relating to chronic gout, non-surgical interventions that include foot orthoses and footwear relating to falls prevention and rheumatoid arthritis, mechanical properties of soft tissue in the foot.

Do textured surfaces affect postural stability and lower limb muscle activity in healthy older adults?

VOLUNTEERS WANTED!!!

Falling is one of the most serious threats to the health and well being of older people. Falls are a major cause of disability and the leading cause of mortality due to injury in older people aged over 75 years in the UK, more than 400,000 older people require A&E attendance and up to 14,000 die annually due to hip fractures. **This PhD study will investigate whether textured surfaces can help in the prevention of falls by way of altering balance?**

What would be involved?

- You would be required to attend the Physiology Lab at the University of Teesside for approximately 45 to 60 minutes on one occasion, to conduct the following tests whilst standing barefoot on the textured surfaces:
    1. Standing quietly with your eyes open for 30 seconds
    2. Standing quietly with your eyes closed for 30 seconds
    3. Standing up from a seated position without using your hands

Who can participate?

- To be eligible to take part in this study you **must be 70 years old or over**
- It is also essential that you **DO NOT** have:
    - Parkinson's Disease, Multiple Sclerosis, or have had a stroke
    - an inner ear disorder (which affects your balance e.g. labyrinthitis)
    - poor sensation on the soles of your feet
    - are unable to walk 10 metres/11 yds unassisted
    - are unable to stand up from a chair without using your hands.

What will you receive?

All participants receive **refreshments** and a **£10 High Street Voucher**. Taxis will be arranged and paid for by the University to bring you to and from your appointment, which will be scheduled at a time convenient to you.

Contact: Anna Hatton (PhD Student, University of Teesside, Middlesbrough)

Tel: 01642 342934 (office), 07952 158930 (mobile), 01642 645330 (home)

UNIVERSITY OF TEESSIDE
SCHOOL OF HEALTH & SOCIAL CARE

PHASE 2 PARTICIPANT INFORMATION LEAFLET

Title of Study: Do textured surfaces affect postural stability and lower limb muscle activity in healthy older adults?

Invitation:

You are being invited to take part in a research project. Before you decide to participate it is important that you understand why the research is being undertaken and what your participation in the study will involve. Please read the following information carefully. You may wish to discuss the matter with others before making your decision. The researcher will be happy to answer any questions you may have about the study, before you decide whether to participate or not.

What is the purpose of the study?

The purpose of the study is to assess the effects of a variety of textured surfaces on leg muscle activity and balance control in healthy adults. This information is important as it may help to identify future ways in which falls prevention could be addressed.

Why have I been chosen?

You have been chosen to participate in this study as you have voluntarily responded to the recruitment poster requesting healthy older adults. In accordance with the exclusion criteria listed on the participant information leaflet, you **do not**: have Parkinson's Disease, Multiple Sclerosis or have had a Stroke, a lack of sensation on the soles of your feet, an inner ear disorder, are unable to walk 10 metres unassisted and/or stand up independently from a chair without using your hands.

Do I have to take part?

It is your decision whether to participate or not in the study. If you do decide to participate you will be asked to sign a consent form. You will be free to withdraw from the study at any time without giving a reason.

What will happen to me if I take part?

After consent is given, you will be asked to attend the Physiology Laboratory at the University of Teesside for approximately 45 minutes on one occasion. You will be asked to remove all footwear, socks/tights and to either wear loose fitting shorts, a skirt, or roll up your trouser leg for this duration. Small areas of skin on your dominant leg will need to be cleaned using isopropyl alcohol swabs. For optimal electrode-skin contact, these areas on your thigh and lower limb may also need to be shaved using an electric razor, before electrodes can be attached. These procedures will be pain-free and will not cause any discomfort. Recording surface electrodes will then be placed over the muscles of interest at specified sites whilst you are standing. Gel will be applied to the electrodes, before they are fixed to your skin with strips of hypoallergenic micropore tape. You may feel a cold sensation as the gel makes contact with your skin.

What do I have to do?

Initially, you will be asked to perform some simple movements of your leg against a moderate resistance, provided manually by the researcher. This ensures the electrodes are positioned correctly over the muscles. You will be instructed to stand barefoot for 30 seconds with your eyes open and then closed. The researcher will announce when testing has begun and when it is over.

What are the possible disadvantages and risks of taking part?

There are no serious side effects or risks anticipated for participants taking part in the study. It is possible that you could have a minor allergic reaction to the hypoallergenic tape used on your leg. If you know of any allergies that you could have to the tape, please bring this to the attention of the researcher. The area where the electrodes are to be placed may need to be shaved. For safety purposes the researcher, who is a qualified physiotherapist, will remain at your side throughout the testing procedures. Any problems should be reported to the Director of Studies, Dr. John Dixon on (01642) 384125

What if something goes wrong?

All parties in the study are covered by the university's indemnity insurance.

What about confidentiality?

All information collected about you during the study will be confidential and will be securely stored and locked away in a fireproof cabinet based at the Centre for Rehabilitation Sciences. Your anonymity will be maintained at all times.

What will happen to the results of the study?

The results of the study will be written up and submitted towards an MPhil/PhD postgraduate degree at the University of Teesside. You will not be identified in any report or publication of the study results. All data will be destroyed at the end of the study.

Who has organised and reviewed the study?

This study is being carried out as part of a research thesis for an MPhil/PhD postgraduate degree at the University of Teesside. Ethical approval has been sought from the School of Health and Social Care Research Ethics Committee, University of Teesside.

Thank you for taking time to read this information leaflet.

For further information please contact: Anna Hatton or Dr John Dixon,
Centre for Rehabilitation Sciences,
School of Health & Social Care,
University of Teesside,
Middlesbrough, TS1 3BA.
Tel: (01642) 342934/384125

Participant No:

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PHASE 2 PARTICIPANT DETAILS

1. **Date of Birth**:
2. **Gender** *(delete as applicable):* **MALE / FEMALE**
3. **Weight** *(kg)*:
4. **Height** *(cm)*:
   Knee Height *(cm):*
   Eye Height *(cm):*

5. **In your own words, describe your *current* general health and fitness. Include details regarding any surgery you have undergone recently *(within the last 12 months)*:**

____

____

7. **Exclusion Criteria** *(please tick the box below if any one or more of the following statements apply to you):*

- You have Parkinson's disease or multiple sclerosis.
- You have had a stroke.
- You have a lack of sensation on the soles of your feet.
- You have an inner ear disorder.
- You require assistance to walk 10 metres/11 yards.
- You are unable to stand up and/or sit down independently from a chair, without using your hands

Please tick here if any one or more of the statements above apply to you: ☐

Participant No:

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7. **Visual Problems** *(delete as applicable):*

- You currently wear glasses and/or contact lenses: **YES / NO**

8. **Foot Posture Index:**

	Factor	**Plane**	**Score**	
			Left -2 to +2	**Right -2 to +2**
Rearfoot	Talor head palpation	Transverse		
	Curves above and below lateral malleolus	Frontal/Transverse		
	Inversion/eversion of the calcaneous	Frontal		
Forefoot	Prominence in the region of the TNJ	Transverse		
	Congruence of the medial longitudinal arch	Sagittal		
	Abduction/adduction forefoot on rearfoot	Transverse		
		Total		
		Foot Posture		

Participant No:

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9. **Monofilament Testing** *(Maximum x3 applications per site per foot):*

Location	Response (Right Foot)	Response (Left Foot)
Between the First and Second Metatarsals (Dorsal Surface)		
Great Toe (Plantar Surface)		
First Metatarsal Head (Plantar Surface)		
Fifth Metatarsal Head (Plantar Surface)		
Heel		
	Insensate / Sensate	**Insensate / Sensate**

Participant No:

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10. **Mini Mental State Examination:**

Orientation	**Score**
1. What is the: • year • date • day • season • month	**(5)**
2. Where are we: • country • county • town • university • floor	**(5)**
Registration	**Score**
3. Name/repeat the following 3 objects: • brush • kettle • flag	**(3)**
Attention and Calculation	**Score**
4. Spell 'world' backwards	**(5)**
Recall	**Score**
5. Ask for the 3 objects repeated above: • brush • kettle • flag	**(3)**
Language	**Score**
6.	
• Name a pencil and a watch	**(2)**
• Repeat the following "No ifs, ands, or buts"	**(1)**
• Follow a three-stage command: 'Take a piece of paper in your right hand, fold it in half, and put it on the floor.'	**(3)**
• Read and obey the following: '**CLOSE YOUR EYES**'	**(1)**
• Write a sentence	**(1)**
• Copy a design	**(1)**
Total	**(30)**

APPENDIX 18: Descriptive statistics for postural sway variables during quiet standing with eyes open over 30 seconds in healthy older adults (n=50)

	Control			**Texture 1**			**Texture 2**		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	5.4 (1.4)	5.0	1.8	5.3 (1.5)	5.2	1.9	5.7 (1.6)	5.4	2.3
AP range (mm)	33.7 (10.5)	31.7	9.1	35.9 (10.2)	36.0	15.7	35.3 (9.4)	33.4	15.6
ML SD (mm)	3.4 (1.1)	3.1	1.8	3.4 (1.2)	3.1	1.6	3.3 (1.0)	3.3	1.7
ML range (mm)	23.9 (8.0)	23.0	10.5	24.7 (8.0)	24.5	9.8	24.2 (9.0)	22.8	9.2
CoP velocity (mm s^{-1})	13.5 (4.2)	12.3	5.2	13.3 (3.6)	13.2	4.3	14.8 (7.6)	12.7	4.6

APPENDIX 19: Descriptive statistics for postural sway variables during quiet standing with eyes closed over 30 seconds in healthy older adults (n=50)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	6.1 (1.9)	5.9	2.4	5.6 (1.6)	5.4	1.9	5.8 (1.8)	5.2	2.4
AP range (mm)	40.4 (13.1)	38.1	23.0	36.3 (11.1)	34.5	18.1	39.2 (12.4)	39.7	16.8
ML SD (mm)	3.9 (1.5)	4.0	2.0	3.6 (1.5)	3.1	1.9	3.8 (1.5)	3.3	2.6
ML range (mm)	27.5 (9.8)	25.4	10.7	25.6 (8.8)	23.8	11.6	27.0 (9.7)	25.2	13.1
CoP velocity (mm s^{-1})	19.3 (7.3)	17.5	9.3	18.4 (6.4)	17.3	8.1	21.2 (8.9)	19.1	9.6

APPENDIX 20: Descriptive statistics for postural sway variables during quiet standing with eyes open over the first 10 seconds in healthy older adults (n=50)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	5.0 (1.6)	4.9	1.6	5.5 (1.7)	5.2	2.4	5.4 (1.7)	4.9	2.3
AP range (mm)	28.2 (10.0)	25.0	10.3	31.4 (10.6)	28.3	15.3	30.2 (9.1)	28.7	11.3
ML SD (mm)	3.5 (1.3)	3.2	1.9	3.6 (1.4)	3.6	1.9	3.6 (1.3)	3.4	1.6
ML range (mm)	21.8 (7.5)	20.2	11.8	22.9 (8.4)	21.9	9.9	22.6 (8.8)	20.7	10.6
CoP velocity (mm s^{-1})	17.4 (5.9)	16.0	7.5	17.8 (5.3)	16.9	6.7	19.2 (8.9)	17.6	7.2

APPENDIX 21: Descriptive statistics for postural sway variables during quiet standing with eyes closed over the first 10 seconds in healthy older adults (n=50)

	Control			Texture 1			Texture 2		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	6.4 (2.7)	6.0	3.2	5.7 (1.8)	5.5	2.6	6.3 (1.9)	6.1	3.2
AP range (mm)	35.8 (14.1)	32.8	18.0	31.9 (11.0)	31.1	14.8	35.6 (12.3)	34.8	17.7
ML SD (mm)	3.9 (1.2)	3.8	1.5	3.7 (1.3)	3.6	1.8	4.0 (1.6)	3.9	2.0
ML range (mm)	24.4 (9.5)	21.5	10.0	23.3 (8.8)	22.7	9.7	24.7 (9.3)	23.4	12.8
CoP velocity (mm s^{-1})	25.9 (10.7)	23.6	12.6	24.0 (7.9)	22.1	12.8	28.3 (10.6)	26.1	12.6

APPENDIX 22: Descriptive statistics for postural sway variables during quiet standing with eyes open over the latter 20 seconds in healthy older adults (n=50)

	Control			**Texture 1**			**Texture 2**		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	4.2 (1.4)	4.2	2.3	4.0 (1.2)	3.9	1.7	4.2 (1.4)	4.1	1.6
AP range (mm)	22.9 (6.6)	22.9	10.9	22.3 (6.1)	22.4	8.4	22.7 (6.1)	21.5	6.8
ML SD (mm)	2.7 (0.9)	2.4	1.1	2.6 (1.0)	2.4	1.3	2.6 (0.8)	2.4	1.5
ML range (mm)	16.7 (5.4)	15.5	6.1	15.9 (4.3)	15.5	5.7	15.8 (4.1)	15.9	6.8
CoP velocity (mm s^{-1})	11.6 (3.7)	10.7	4.7	11.0 (3.2)	11.1	4.0	12.6 (7.3)	10.4	4.4

APPENDIX 23: Descriptive statistics for postural sway variables during quiet standing with eyes closed over the latter 20 seconds in healthy older adults (n=50)

	Control			**Texture 1**			**Texture 2**		
	Mean (SD)	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
AP SD (mm)	4.9 (1.7)	4.7	1.9	4.6 (1.6)	4.0	1.9	4.6 (1.5)	4.3	1.7
AP range (mm)	27.3 (9.1)	26.0	11.8	25.8 (7.6)	24.6	9.7	26.4 (8.8)	24.8	10.9
ML SD (mm)	3.0 (1.3)	2.6	1.8	2.7 (1.0)	2.4	1.6	2.9 (1.1)	2.6	1.5
ML range (mm)	18.5 (7.3)	16.8	8.3	16.7 (4.7)	15.3	7.7	17.4 (5.5)	16.5	6.3
CoP velocity (mm s^{-1})	16.0 (6.7)	14.1	9.0	15.6 (6.1)	14.1	7.6	17.6 (8.7)	15.5	8.2

APPENDIX 24: Descriptive statistics for lower limb EMG activity (μV) during quiet standing with eyes open over 30 seconds in healthy older adults (n=50)

		Control			Texture 1			Texture 2		
	n =	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
Rectus Femoris	50	12.4 (4.1)	10.1	5.0	12.3 (4.1)	10.0	5.1	12.6 (4.9)	10.0	5.6
Vastus Lateralis	46	14.5 (15.6)	9.6	17.5	15.1 (16.9)	9.4	20.4	14.8 (17.0)	8.2	19.5
Biceps Femoris	46	6.0 (10.4)	2.0	2.2	6.4 (9.9)	2.1	5.5	8.3 (15.5)	1.9	3.5
Tibialis Anterior	50	3.4 (3.4)	1.8	2.7	3.1 (2.9)	2.0	1.1	3.6 (3.7)	1.9	2.2
Medial Gastrocnemius	49	6.9 (5.7)	5.4	5.5	7.9 (8.8)	5.5	6.4	7.6 (7.0)	6.3	7.2

APPENDIX 25: Descriptive statistics for lower limb EMG activity (μV) during quiet standing with eyes closed over 30 seconds in healthy older adults (n=24)

		Control			**Texture 1**			**Texture 2**		
	n =	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
Rectus Femoris	50	13.5 (4.9)	11.0	7.4	13.0 (5.0)	10.6	5.8	13.3 (4.8)	10.7	7.2
Vastus Lateralis	46	16.5 (15.2)	12.0	23.4	16.0 (17.8)	10.9	21.5	17.2 (17.7)	11.9	24.0
Biceps Femoris	48	6.4 (9.9)	2.2	3.3	6.2 (9.3)	2.0	2.4	6.0 (9.6)	2.1	2.9
Tibialis Anterior	50	6.4 (8.6)	2.1	4.9	5.2 (7.3)	2.3	5.3	6.0 (7.7)	2.2	3.1
Medial Gastrocnemius	49	8.9 (7.4)	6.1	10.1	8.8 (8.2)	6.5	8.1	9.3 (7.5)	7.3	10.5

APPENDIX 26: Descriptive statistics for lower limb EMG activity (μV) during quiet standing with eyes closed over the first 10 seconds in healthy older adults

		Control			Texture 1			Texture 2		
	n =	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
Rectus Femoris	50	29.0 (5.3)	26.2	7.4	28.6 (5.9)	26.0	6.2	28.7 (5.2)	25.9	6.0
Vastus Lateralis	46	17.5 (16.7)	11.6	24.3	17.9 (21.5)	11.4	21.8	18.2 (18.6)	12.7	28.1
Biceps Femoris	48	6.5 (10.2)	2.2	4.0	6.4 (10.6)	2.0	2.7	6.8 (11.2)	2.1	3.3
Tibialis Anterior	50	6.7 (9.7)	2.2	4.3	5.6 (8.1)	2.2	5.0	6.1 (8.9)	2.1	4.1
Medial Gastrocnemius	49	8.8 (7.7)	5.9	9.0	8.3 (7.2)	6.2	7.9	8.8 (6.9)	7.0	8.0

APPENDIX 27: Descriptive statistics for lower limb EMG activity (μV) during quiet standing with eyes closed over the latter 20 seconds in healthy older adults

		Control			Texture 1			Texture 2		
	n =	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**	**Mean (SD)**	**Median**	**IQR**
Rectus Femoris	50	5.8 (4.8)	3.5	7.1	5.1 (4.6)	2.9	5.7	5.5 (4.6)	2.7	7.3
Vastus Lateralis	46	16.0 (14.8)	12.2	24.1	15.1 (16.3)	10.5	21.7	16.7 (17.4)	11.9	21.4
Biceps Femoris	48	6.4 (10.0)	2.2	3.0	6.2 (8.9)	2.0	2.3	5.6 (8.9)	2.1	2.8
Tibialis Anterior	50	6.3 (8.8)	2.1	3.6	5.1 (7.4)	2.1	3.3	5.5 (7.2)	2.1	2.5
Medial Gastrocnemius	49	9.0 (7.5)	6.2	10.9	9.1 (8.9)	7.0	9.2	9.6 (7.9)	7.3	11.0

APPENDIX 28: Mean postural sway data at baseline for healthy older males (n=21) and females (n=29) during quiet standing on the control surface. Sway data highlighted in red indicates where females showed a trend for greater mean sway than males.

Sway Variable	Gender	Eyes Open			Eyes Closed		
		10 seconds	20 seconds	30 seconds	10 seconds	20 seconds	30 seconds
		Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
AP SD	Males	5.3 (1.8)	4.5 (1.5)	5.7 (1.5)	6.9 (3.0)	5.0 (1.6)	6.3 (1.9)
	Females	4.8 (1.5)	4.0 (1.3)	5.1 (1.3)	6.1 (2.4)	4.8 (1.8)	5.9 (1.8)
AP Range	Males	29.4 (10.6)	23.5 (7.4)	36.0 (12.2)	40.0 (15.5)	27.7 (7.8)	43.8 (13.6)
	Females	27.3 (9.6)	22.4 (6.1)	32.1 (9.0)	32.7 (12.4)	26.9 (10.1)	38.0 (12.3)
ML SD	Males	3.5 (1.2)	2.6 (1.0)	3.3 (1.1)	4.2 (1.2)	3.1 (1.3)	4.4 (1.5)
	Females	3.5 (1.4)	**2.7 (0.8)**	**3.4 (1.1)**	3.7 (1.2)	2.9 (1.3)	3.6 (1.3)
ML Range	Males	22.9 (8.5)	15.8 (4.7)	24.7 (8.7)	27.1 (11.2)	18.1 (6.3)	30.9 (11.1)
	Females	21.0 (6.8)	**17.4 (5.9)**	23.3 (7.5)	22.4 (7.7)	**18.7 (8.0)**	25.1 (8.1)
CoP Velocity	Males	17.4 (4.9)	11.4 (2.9)	13.4 (3.2)	29.9 (9.8)	17.9 (7.2)	21.9 (7.6)
	Females	17.4 (6.7)	**11.6 (4.3)**	**13.6 (4.8)**	23.0 (10.5)	14.6 (6.1)	17.4 (6.6)

APPENDIX 29: Exploratory sub-group analysis for the effects of textured surfaces on postural sway parameters in healthy older males (n=21) and females (n=29) during quiet standing with eyes open and eyes closed

		Eyes Open			Eyes Closed		
	Gender	10 seconds ANOVA *p*=	20 seconds ANOVA *p*=	30 seconds ANOVA *p*=	10 seconds ANOVA *p*=	20 seconds ANOVA *p*=	30 seconds ANOVA *p*=
AP SD (mm)	Males	0.366	0.786	0.976	0.370 gg	0.738 gg	0.766
	Females	0.180	0.898	**0.014***	**0.038***	0.238	**0.053***
AP range (mm)	Males	0.146	0.992	0.394	0.362 gg	0.841	0.629 gg
	Females	0.343	0.577	0.168 gg	**0.053***	0.428	**0.038***
ML SD (mm)	Males	0.082	0.177 gg	0.128	0.356	0.137	0.131
	Females	0.763	0.103 gg	0.729	0.708	0.081 gg	0.154
ML range (mm)	Males	0.268	0.106	0.179	0.665	0.487	0.376
	Females	0.763	0.087 gg	0.780	0.688	**0.051 gg***	0.491
CoP velocity (mm s^{-1})	Males	0.272 gg	0.250 gg	0.274 gg	**0.047***	0.922	0.419
	Females	0.171	0.444	0.256	**0.001***	0.058 gg	**0.007***

* p<0.05, gg = Greenhouse-Geisser correction for sphericity

APPENDIX 30: Exploratory sub-group analysis for the effects of textured surfaces on lower limb muscle activity in healthy older males (n=21) and females (n=29) during quiet standing with eyes closed

		Eyes Closed		
	Gender	10 seconds ANOVA *p=*	20 seconds ANOVA *p=*	30 seconds ANOVA *p=*
Rectus Femoris	Males	0.573 gg	0.575 gg	0.572 gg
	Females	0.229	0.069	0.097
Vastus Lateralis	Males	0.820 gg	0.459 gg	0.585 gg
	Females	0.669	0.899	0.862
Biceps Femoris	Males	0.842	0.471	0.595
	Females	0.073	0.120 gg	0.090
Tibialis Anterior	Males	0.284 gg	0.481 gg	0.407 gg
	Females	0.436	0.318	0.126 gg
Medial Gastrocnemius	Males	0.532 gg	0.731	0.789
	Females	0.159	**0.049***	0.057

* p<0.05, gg = Greenhouse-Geisser correction for sphericity

American College of Sports Medicine 55th Annual Meeting
28th-31st May 2008, Indianapolis, Indiana

DO TEXTURED SURFACES AFFECT POSTURAL STABILITY AND LOWER LIMB MUSCLE ACTIVITY IN YOUNG ASYMPTOMATIC ADULTS?

Anna Hatton, John Dixon, Keith Rome, Denis Martin, David Hodgson

Shoe insoles may enhance functional ability. Vibrating insoles have reduced postural sway in healthy young adults during quiet standing balance (QSB) and textured insoles have altered lower limb muscle activity during dynamic tasks. Enhanced sensory input to the soles of the feet may influence postural stability, with the underlying mechanism being altered lower limb muscle activity.

PURPOSE: This study investigated whether textured surfaces alter anterior-posterior (AP) and mediolateral (ML) postural sway and lower limb EMG intensity during QSB.

METHODS: 24 asymptomatic adults (17F, 7M), mean (SD) age 27.5 (7.9) years, conducted bilateral QSB tests (30 seconds, eyes open) under randomised conditions, by covering a force plate with three textured surfaces differing only in their pattern of indentation. AP and ML range and standard deviation (SD) (mm) were extracted and surface EMG amplitude (μV) recorded from eight dominant lower limb muscles. All data were sampled at a rate of 1000Hz. EMG signals were high pass filtered at 20Hz and averaged using a 30ms root-mean-square window. Differences in postural sway and EMG activity between textured surfaces were evaluated by examining the mean differences between conditions.

RESULTS:

Table 1: Mean (SD) of postural sway variables (mm).

Condition	AP Range	AP SD	ML Range	ML SD
Control	24.5 (5.7)	4.4 (1.4)	18.2 (3.4)	3.0 (0.7)
Texture 1	23.1 (3.6)	4.1 (0.9)	18.2 (3.3)	3.1 (0.7)
Texture 2	24.8 (3.0)	4.6 (0.8)	18.3 (2.9)	3.0 (0.6)

The textured surfaces had no substantial effect on postural sway or EMG amplitude in all eight lower limb muscles.

CONCLUSIONS: Textured surfaces do not substantially affect short-term postural sway and lower limb EMG in healthy young adults. Further work must determine whether textured surfaces; (i) alter postural sway with accompanying changes in lower limb EMG during dynamic tasks and (ii) ameliorate age-related impairments in postural stability.

9th International Falls and Postural Stability Conference
19th September 2008, York, UK

DO TEXTURED SURFACES AFFECT POSTURAL STABILITY DURING QUIET STANDING BALANCE IN HEALTHY OLDER ADULTS?

Hatton AL, Dixon J, Martin D, Rome K

Introduction: Immediately following sit-to-stand (STS), the centre of gravity must re-stabilise after a potentially de-stabilising functional movement: one which may be associated with falling in older adults. Footwear interventions, including textured insoles, may help control standing posture by enhancing sensory input to the soles of feet. This study aimed to investigate whether textured surfaces alter postural sway during quiet standing balance (QSB) in healthy older adults over 10, 20 and 30 seconds.

Methodology: 50 healthy older adults (age mean 75.1,SD 5.0 years; 29 female) conducted bilateral QSB tests (30 seconds) following STS, with eyes open (EO) and closed (EC) under randomised conditions. Two textured surfaces of differing indentations (T1, T2) and a smooth surface as control were laid over a force plate. Anterior-posterior (AP) and mediolateral (ML) range and standard deviation (mm) were extracted over the initial 10, latter 20, and overall 30 seconds of QSB.

Results: Repeated measures ANOVA showed a statistically significant difference in ML Range with EC ($p=0.03$) over the latter 20 seconds of QSB. Pairwise comparison identified a mean difference of -1.7 (95% CI -3.3 to -0.2) mm between C and T1. No significant differences were observed for any other postural sway variables with EO or EC, over all the three time frames ($p>0.05$).

Conclusion: With EC T1 significantly reduced ML sway following an initial period of re-stabilisation, in comparison to control levels. With EO textured surfaces did not significantly alter the magnitude of postural sway in healthy older adults during QSB. The evidence of an effect of one of the textured surfaces on ML sway supports the possibility of this as a therapeutic option.

3rd Australian and New Zealand Falls Prevention Conference
12th-14th October 2008, Melbourne, Australia

THE EFFECT OF TEXTURED SURFACES ON POSTURAL STABILITY AND LOWER LIMB MUSCLE ACTIVITY IN HEALTHY OLDER ADULTS

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Aim: Footwear interventions may improve balance control in older adults. Textured insoles may help ameliorate age-related postural instability by enhancing sensory input to the soles of the feet: this has not been investigated in older adults. This study aimed to determine whether textured surfaces alter the magnitude of postural sway and intensity of lower limb muscle activation during quiet standing balance (QSB) in healthy older adults.

Method: 50 healthy older adults (age mean 75.1, SD 5.0 years, 28 female), conducted bilateral QSB tests (30 seconds) with eyes open and closed under randomised conditions. Two textured surfaces of differing indentations (T1, T2) and a smooth surface as control were laid over a force plate. Centre of pressure (CoP) velocity (mm s^{-1}), anterior-posterior and mediolateral SD and range (mm) were extracted and EMG amplitude (μV) recorded from five dominant lower limb muscles.

Results: Repeated measures ANOVA showed a statistically significant difference in CoP velocity with eyes closed ($p=0.005$). Pairwise comparison identified a mean difference of 2.8 (95% CI: 0.5 to 5.0) mm s^{-1} between T1 and T2. No significant differences were reported for any other postural sway variable or EMG amplitude in all 5 lower limb muscles ($p>0.05$).

Conclusion: Textured surfaces did not significantly change the magnitude of postural sway or amplitude of lower limb EMG from control levels during QSB in healthy older adults with eyes open or closed. However, in the absence of visual stimulation, the two textured conditions appear to have opposite effects, suggesting further research is necessary.

British Geriatrics Society Autumn Meeting

12th – 14th November 2008, Birmingham, UK

THE EFFECT OF TEXTURED SURFACES ON POSTURAL STABILITY AND LOWER LIMB MUSCLE ACTIVITY IN HEALTHY OLDER ADULTS

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INTRODUCTION: Footwear interventions may improve balance control in older adults. Textured insoles may help ameliorate age-related postural instability by enhancing sensory input to the soles of the feet: this has not been investigated in older adults. This study aimed to determine whether textured surfaces alter postural sway and muscle activation during quiet standing balance (QSB) in older adults.

METHODS: 50 healthy older adults (mean age 75.1 ± 5.0 years), conducted bilateral QSB tests (30 seconds) with eyes open and closed under randomised conditions. Three textured surfaces of differing indentations (C, T1, T2) were laid over a force plate. Centre of pressure (CoP) velocity (mm s^{-1}), anterior-posterior and mediolateral range and standard deviation (mm) were extracted and EMG amplitude (μV) recorded from five dominant lower limb muscles.

RESULTS: Repeated measures ANOVA showed a statistically significant difference in CoP velocity with eyes closed ($p<0.05$). Post-hoc analysis identified a mean difference of 2.8 (95% CI: 0.5 to 5.0) mm s^{-1} between T1 and T2. No significant differences were reported for any other postural sway variable or EMG amplitude in all 5 lower limb muscles ($p>0.05$).

CONCLUSIONS: Textured surfaces did not significantly change the magnitude of postural sway or amplitude of lower limb EMG from control levels during QSB in healthy older adults. However, in the absence of visual stimulation, the two textured conditions appear to have opposite effects, suggesting further research is necessary.

Physiotherapy Research Society Spring Meeting
7th May 2009, Glasgow, Scotland

TEXTURED SURFACES ALTER MEDIOLATERAL POSTURAL SWAY IN HEALTHY OLDER PEOPLE

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Purpose: To investigate whether textured surfaces alter postural sway variables and lower limb muscle activity in healthy older adults during quiet standing.

Relevance: Previous studies have shown balance can be improved by surfaces with texture compared to those that are smooth. Textured footwear interventions can also influence lower limb muscle activity, possibly by providing enhanced sensory input. No study has investigated the effect of texture on postural stability and lower limb muscle activity simultaneously in older people.

Subjects: 50 healthy older adults (29 female); mean (SD) age 75.1 (5.0) were investigated.

Methods: Ethical approval was granted by the School of Health & Social Care Research Governance and Ethics Committee at the University of Teesside. Anterior-posterior and mediolateral sway variables and lower limb EMG intensity in five muscles were collected synchronously over 20 seconds of bipedal standing, with eyes open and closed, under three randomised conditions; Control, Texture 1, Texture 2.

Analysis: One-way repeated measures ANOVA

Results: Texture 1 significantly reduced mediolateral sway range compared to control, when standing quietly with eyes closed. Mean (95% confidence interval) difference between these surfaces was 1.7 (0.2 to 3.3) mm, (p=0.03). No such effects were seen in any other postural sway variable or lower limb EMG activity, for either visual condition.

Discussion: Textured surfaces can improve standing balance in healthy older people. Alterations in balance were not accompanied by changes in lower limb muscle activity. The potential to help reduce the risk of falling in older people using textured surfaces requires further investigation.

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